ON THE IMPLEMENTATION OF OPENSIM: APPLICATIONS OF
MARKER-BASED AND INERTIAL MEASUREMENT UNIT BASED SYSTEMS

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The gait analysis system of METU Mechanical Engineering Department was established in the mid 1990s. The measurement technique is based on capturing the marker locations by means of video-cameras. The software packages employed are responsible for processing of video-camera images to obtain the 3-D positions of markers attached to specific locations of body and inverse kinematics and dynamics to reveal the joint angles, forces and moments during a gait.

The first goal of the study is to analyze alternative methods of obtaining joint kinematic parameters from the marker locations by implementing different rotation angle sequences. The results indicate that, for a typical gait, different rotation sequences do not cause severe differences in the joint angles. Additionally, an open-source software, OpenSim, is utilized to process marker data, which provides a simulation environment for the existing system. Different skeletal models are created and critically analyzed with the aid of OpenSim. The last part of the study is devoted to the development of a new system using inertial measurement units. Each inertial measurement
unit attached to body segments is responsible for extracting the orientation of the body in 3-D. The joint kinematic parameters and simulation are handled employing OpenSim, through subroutines written using its libraries.

Keywords: Gait Analysis, Euler Angles, OpenSim, Inertial Measurement Unit
ÖZ

OPENSİM YAZILIMININ UYGULANMASI ÜZERİNE: İŞARETLEYİCİLİ VE ATALETSEL ÖLÇÜN BİRİMİ SİSTEMLERİN UYGULAMALARI

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Çalışmanın ilk amacı, farklı dönme açısı dizileri uygulayarak, eklem kinematik parametrelerinin işaretleyici konumlarından çıkarım yöntemlerini analiz etmektir. Sonuçlar, tipik bir yürüyüş için farklı dönme dizilerinin eklem açılarında ciddi farklılıklarla neden olmadığını göstermektedir. Ayrıca, bu tez sayesinde sisteme bir yürüyüş andırın ortamı sağlanmıştır. Açık kaynak kodlu bir yazılım olan OpenSim, sistemin işaretleyici verilerini işlemek için kullanılır. Farklı iskelet modelleri oluşturulmuş ve eleştirel olarak analiz edilmiştir. Çalışmanın son kısmı, atalet ölçüm birimlerini kullanan yeni bir sisteme öncülük etmektedir. Her bir vücut uzvuna bağlı her birim,
uzvun duruşunu 3B olarak çıkarmaktan sorumludur. Eklem kinematik parametreleri ve andırım, OpenSim kütüphaneleri kullanılarak yazılan alt yordamlar aracılığıyla gerçekleştirilir.

Anahtar Kelimeler: YürüyüŞ Analizi, Euler Açıları, OpenSim, Ataletsel Ölçüm Birimi
To my family
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It has been a long time since human walking and locomotion attracted the attention of scientists. Starting with the observations by specialized personnel, the technological and computational advances bring the study of the human walking to a level of high accuracy and repeatability by supplying quantitative and objective data with the measurements and their interpretation. From simple observations to automated computer techniques used, the subject matter is called gait analysis. Biomechanics, in general, is an area involving the application of classical mechanics to biological systems and investigates the behavior of bodies under the action of forces. Gait analysis is a special area of biomechanics which considers the style of human walking. The first step in gait analysis is to extract information related to the kinematic and kinetic parameters of the bodies involved from the measured 3-D positions of specific landmarks on several segments of the human body and ground reaction forces acting on the feet. Kinematic parameters, specifically the joint angles, help to describe the human gait quantitatively. Further, they can be used for the diagnosis and treatment planning of pathological forms of human gait. Whether the physical treatment or the surgery has positive effects on the mechanics of gait is answered by comparing the walking, or any other kind of movement, of the subject before and after the corresponding treatment.
1.1 Motivation

Biomechanics laboratory at METU houses the first-ever gait analysis system in Turkey, called KISS\textsuperscript{1}. The system uses markers, video cameras and force plates to capture and analyze human movement. From the establishment up to now, there have been many research and clinical applications carried out in the laboratory, as listed below:

- Güler established the laboratory setup including hardware, software and biomechanical model \cite{1}.
- Shafiq established the 3-D marker position extraction from multiple cameras \cite{2}.
- Karpat assessed the capability of tracking of marker trajectories of the system \cite{3}.
- Afşar evaluated the marker errors caused by skin movement \cite{4}.
- Söylemez documented the biomechanical model and investigated the reliability and repeatability of the system \cite{5}.
- Civek investigated the kinematic results of the laboratory setup based on a commercial gait analysis system \cite{6}.
- Kafalı assessed sensitivity and compatibility of the analysis protocol by varying the experimental methodology \cite{7}.
- Erer proposed adaptive version of Butterworth filter used in smoothing of the marker trajectories and implemented kinetic formulation, which was originally on Delphi, on Matlab \cite{8}.

Based on the previous works briefly summarized above, the initially adapted model for the joint angle calculation was used intensively without any change and investigation. The first two parts of this thesis is concerned with analyzing the effects of different sequences in joint angle calculations, and introducing simulation capability to the existing system. The last part of this thesis is devoted to measurements with a

\textsuperscript{1} Abbreviated from Kas İŞkelet Sistemi in Turkish (by S. Turgut Tümer) and KInematic Support System in English (by Necip Berme)
different technique, specifically employing inertial measurement units to capture 3-D position and orientation of body segments.

1.2 Scope of the Research

The first part of this research investigates different mathematical models using Euler/Cardan rotation angle sequences and compares the results with the KISS system.

The second part of the thesis is concerned with introducing a simulation environment utilizing OpenSim. The measured quantities of KISS system are transformed into the OpenSim environment input files and existing musculoskeletal models are employed. The kinematic results are investigated and compared with those of the KISS system. Different models and different marker sets are generated to investigate their effects. In addition to joint angles, the translations in the joints are quantified, which is not existing and therefore not studied and documented in the present KISS system. The marker errors, defined as the difference between experimental marker locations and those of musculoskeletal model markers of OpenSim, are examined by altering the model and marker sets.

The last part of the thesis is related to a different motion capture technique. The original system having the markers and video-cameras used to measure the spatio-temporal parameters is aimed to be replaced by a new technology. The new system utilizes the inertial measurement units to measure angular velocities, linear accelerations in the bodies attached and the magnetic field of the earth. This new data set is filtered and processed by an algorithm. The last step is to simulate gait with the new data set. Although OpenSim requires markers to simulate gait, the subroutines are written to let OpenSim use the results of the inertial measurement units in simulation.

1.3 Outline of the Thesis

Chapter 2 includes the background information by overviewing the existing gait analysis methods and biomechanical models.
Chapter 3 is a detailed work on the KISS gait analysis system. The investigations and discussions on the effects of different joint angle calculation procedures are presented.

Chapter 4 is devoted to the new simulation environment, OpenSim. The input files for simulation are generated, and different models and marker sets are critically evaluated.

Chapter 5 employs inertial measurement units in the human movement capture and gait analysis.

Chapter 6 contains conclusions of the present study and recommendations for future work.
Gait analysis is a research topic which involves a systematic study of locomotion of a living being, especially the human. The movement includes level walking, running, stair ascend and descend and performing sports activities like playing football, basketball and more. The importance of the study comes from the fact that normal and pathological forms of movement can be distinguished by this analysis. There are different ways of conducting a gait analysis. The earliest method is the subjective visual inspection and evaluation done by a medical doctor. With the advance of technology and the need to store previous gait data, electro-goniometers, electromyography, video camera based systems, inertial measurement units and ultrasonic systems have been used. All instrumented gait analysis systems supply objective and quantitative data. One of the most important studies in the gait analysis area is the computer simulation of the subject gait which helps to conduct further analyses such as answering what-if questions.

Gait analysis protocol is composed of the measurement technique (the most common and widely used method being marker sets in video-camera based systems) and the related biomechanical model to describe the motion of the human body mathematically.

In this chapter first, techniques to measure different aspects of gait will be presented. Then, the video-camera based systems which use passive markers are investigated. Following the marker set definitions, biomechanical models used in literature are discussed in terms of reference frame definitions and solution methods. Finally, open-source simulation environment, OpenSim, is described in relation to its application to gait analysis.
2.1 Gait Analysis Methods

2.1.1 Observational Gait Analysis

Visual observation made by a clinician is the simplest form of the gait analysis. An expert may analyze the gait by the naked eye and detect gait anomalies, if any. With the advance of technology, and need to store gait data to compare the results, the observational gait analysis is no longer sufficient in many cases. For example, a clinician may need to analyze the gait of a subject before and after a specific treatment, which is best achieved with quantitative data.

2.1.2 Camera Based Methods

Either real (passive or active markers) or imaginary (marker-less systems that artificially place markers on the body) markers are used to construct the bodies in 3-D. A body in 3-D space can be expressed by the cartesian coordinates of 3 non-collinear points on it. To successfully define the segments in the lower body, there are different marker sets proposed in the literature which will be discussed in Section [2.2.1].

There are different techniques to find the landmarks on a body in camera based systems. Active markers are light-emitting diodes (LEDs) [9]. The special optoelectronic cameras locate the markers in their own 2-D space by the light emitted from it. The LED markers are illuminated by the power supply and wires which are carried by the subject. The major shortcoming of this system is the distraction of the subject’s gait by carrying some bulky equipment. Also, there are visibility issues like the reflections caused by the markers which may not be detected. These problems make active marker systems less popular in the gait analysis community.

The passive markers are generally solid spheres covered with reflective tape, which are, thus, called retroreflective markers [10]. The illumination is done by the LEDs around each camera lens, and there is no need for the subject to carry unnecessary power source and wiring. The reflected light is captured by the cameras which are generally filtered to infrared spectrum. According to [11], the most common measurement method employed in the gait analysis community is the passive markers. L. Chèze [11] states that position of a marker center in 3-D space can be reconstructed
from the coordinates of the same point in the planes of at least two cameras with a prior calibration. Calibration methods using direct linear transform (DLT) and epipolar geometry is presented by L. Chèze in detail. In brief, the 3-D position of a marker can be determined by at least two 2-D images of camera planes which are not parallel. The point at which orthonormal epipolar lines of camera planes intersects is the 3-D coordinate of the marker. The same principle is used for active markers and for virtual markers in markerless systems as well.

2.1.3 Electro-Goniometer

Electro-goniometer converts the rotary motion into electrical currents. The proximal and distal link segments are strapped with a rigid link, while the joint angle is observed by the electrical output of the device. Due to difficulties in their application and measuring only relative joint angles in a single plane they are not preferred [10].

2.1.4 Inertial Measurement Unit

Inertial measurement unit (IMU) is a device including accelerometer, gyroscope and magnetometer. The linear accelerations of a body in three mutually perpendicular directions is sensed by a 3-axis accelerometer, which makes use of Newton’s Second Law to obtain the accelerations caused by the forces acting on the body. The gyroscope measures the angular velocity of a body from the principle that a body revolving around an axis develops rotational momentum which produces the resistance to change its angular speed vector. Micro electro-mechanical system (MEMS) gyros are tiny devices that include a small mass between the springs which shifts as the angular velocity changes. Additionally, IMU includes a magnetometer that points magnetic north, and this information is used to decrease the inherent error [12]. Application of IMU’s to gait analysis is a relatively new technique.
2.1.5 Markerless Systems

In order to not disturb the motion of subject or in some sports events where attaching anything on the body is not allowed, there are attempts to capture and analyze the human movement without any markers. These systems, called markerless systems, are only dependent on a number of cameras. After calibrating the cameras, the movement is captured by means of special computer vision algorithms. First, background subtraction is applied. One such example, by Sandau et al. [13], uses patch based multiview stereo algorithm to construct 3-D points. The subject specific model is constructed without any joint constraints. The results are comparable to that of marker-based system in the sagittal plane and hip abduction. Another work concerning the joint center location identification was carried out by Prakash et. al. [14]. Using sophisticated image processing methods like silhouette image extraction and image segmentation, comparable results were obtained as compared to marker-based systems.

2.1.6 Additional Measurements

The aforementioned methods measure the kinematics of gait which includes position, velocity, acceleration and time information. The kinetic (or dynamic) part of the analysis deals with the forces, especially the forces and moments at the joints. To obtain joint reaction forces and moments, the ground reaction forces must be known, either measured directly or calculated under some assumptions. Further, muscle activity can be measured to study actuator dynamics of human gait.

The ground reaction is composed of three force components ($F_x$, $F_y$ and $F_z$) along the principal directions and three moment components ($M_x$, $M_y$ and $M_z$) about the principal directions. The force plates (or sometimes platforms) are load cells that can measure from one component (the vertical force) up to the six components described above. Some types include the information of center of pressure, that is, the point of application of net force [10]. The process of measuring the electrical muscle activity is called electromyography (EMG). Since the muscles are in charge of generating and controlling the movement of the body segments, it is beneficial to know and
analyze their role \cite{15}. The musculoskeletal models utilize information from this measurement. The joint forces and moments are the resultants of muscle forces which cannot be analytically calculated from kinetics.

2.2 Biomechanical Model

In this section, biomechanical models will be presented for the camera-based marker systems. The differences in the biomechanical model consist of marker sets, degree of freedom of joints (which is related to the marker sets), reference frame definitions, joint rotation conventions (inverse kinematics step).

2.2.1 Marker Sets

Marker sets are the standard arrangement of markers to perform kinematic analysis \cite{16}. The markers are placed on the pre-defined landmarks on the segments of the human body that will be tracked during the activity to re-construct the segment on a computer in 3-D. Positions of markers are acquired in 3-D global coordinate system of the laboratory. While there must be three non-collinear markers for each segment, there are different placement procedures for the markers in the literature. They are basically either 3-dimensional or 6-dimensional. The name dimensional is given for the allowed degrees of freedom of the segments in the marker sets.

2.2.1.1 3-DOF Marker Set

The most common marker set in the literature is called Helen-Hayes (HH) marker set which is defined by the prior works presented in \cite{17}. Name of the marker set is due to first work conducted in Helen Hayes Hospital. The marker set uses common points to define segments. Kadaba et al. \cite{17} introduce basic assumption which is relative rotation between segments takes place about a fixed point. Thus, the joint between the bodies is a spherical joint and they share a common point called joint center which corresponds to the center of the spherical joint. The name of the marker set comes from the fact that a spherical type joint has only 3 rotational degrees of
freedom. A total of 13 markers (3 on pelvis and 5 on each leg) are used for the lower extremity. The location an orientation of the pelvis is defined by 3 markers attached on it. By defining the hip joint centers in the pelvis, a point common to hip and thigh is obtained. Attaching two markers on thigh and the knowledge of hip joint center (sometimes called center of femoral head) is enough to construct thigh. To construct the next segment, the knee joint center and two additional markers are used. Finally, the foot is defined in 3-D space by ankle joint center and two markers attached on it. The HH marker set is then modified by [18] and [19]. These sets are generally called Modified HH marker set and sometimes as Newington, Davis, Vicon Clinical Manager or Vaughan. The Modified HH marker set is presented in Figure 1. The modification is an additional marker attached on each foot to express its position in 3-D accurately. The set, thus, is composed of 15 markers. The advantage of HH marker set over the 6D sets defined in the following section is that it is simpler due to less number of markers used [20]. The less number of markers makes the experimental preparation procedure easier. Additionally, there is less disturbance on the subject while performing the activity. If the translational degrees of freedom of the joints are ignored, Modified HH marker set is capable of accurately estimating the segment orientations. In [20], Sutherland argued that HH marker set is more applicable to children since there is no risk of markers hitting each other. Kadaba et al. in [17], argued that the number of markers should be minimum but enough to define segments in 3-D. Although the easiness and simplicity of the experimental procedure is taken into consideration in terms of clinical interpretation, reproducing the actual joint movement in detail is an important aspect of research.

2.2.1.2 6-DOF Marker Set

If the joints are considered to have all degrees of freedom, in fact they do, the assumption of common, or permanently coincident, points on each segment fails. Therefore, 6D marker set is required to track each segment independently. The group of markers is used for each segment for independent definition. These groups, called clusters, construct the technical frames for the segments. The most commonly used set is the Cleveland Clinic Marker set [20]. According to [21], the clusters are mounted on rigid base plates which are then strapped or taped to the segments in the early
applications. Some of the models do not use cluster for pelvis and foot. Although Cappozzo et al. in [22] uses marker cluster technique, it contradicts with the advices for marker placement given in [23] by Cappozzo. In [23], relative movement between the marker and underlying bone minimization and sufficiently large distance between the markers are advised. The first advise can be obtained by using anatomical landmark locations since they are less sensitive to skin movement. Since the 6-DOF set is based on marker cluster placed over the skin and far from the bone, there will be more skin movement. The second advice can be met by using large clusters to make distance between markers far, however, this will result in encumbering the subject. Although there are some drawbacks, this method may be employed to eliminate the error due to the 3-DOF joint assumption which affects the distal segment more [16]. In [24], Li et al. showed that the knee joint translations may be in the order of 15 mm.

![Figure 2.1: Modified HH Marker Set (reproduced from [15])](image)

### 2.2.2 Reference Frame Definitions

There are three reference frames used in the gait analysis. The global reference frame is the earth fixed laboratory frame according to which the segments are defined. The establishment of the global frame may vary from laboratory to laboratory. However, this will not affect the results since transformation between two earth fixed frames will be constant for all practical purposes.

The local, or segmental reference frames are defined for each segment. There are
two types of local frames, namely technical and anatomical. The former is used for tracking purposes, while the latter is constructed to extract the information of segment’s orientation with respect to the anatomical planes. Cappozzo et al. in [25] put the following requirements on the local (or segmental) reference frames;

- It should be repeatable both inter and intra-individually
- It should allow determination of suitable axes about the allowed rotations (in 3-DOF or 6-DOF marker sets) or along allowed translations (only in 6-DOF marker set) of the joints.
- It should allow determination of the parameters that will be used in dynamic analysis easily, such as center of mass of the segment
- Description of muscle and ligament line of action should also be considered (when the muscle forces are involved in the analysis)

These can only be assured by the reference frames that is associated with the anatomy. These frames are called anatomical frames and are based on anatomical landmarks. Anatomical landmarks are easy to locate and their reproducibility is assured according to [26], less prone to skin motion according to [27]. On the other hand, the markers are required to be external to the body surfaces due to experimental requirements, such as non-invasive application, visibility to cameras and easy implementation. Reference frame defined by the markers is called technical frame which is in an arbitrary but constant relationship with respect to the anatomical frame. One of the earliest works to define anatomical local frames is the in vitro study by Ruff et al. [28]. The definition is based on landmarks and measurements on the bones. The next and most widely accepted definition of anatomical reference frame is by Cappozzo et al. [22]. Pelvis is composed of three segments, namely the two hip bones and the sacrum. Cappozzo et al. [22] assumed that the relative movement between these three segments is negligible, and following anatomical frames are defined.

- Pelvis frame:
  - Origin is the midpoint of left and right anterior superior iliac spines (LASIS and RASIS).
- z-axis is through ASISs and positive from left to right.
- x-axis lies in the quasi transverse plane defined by ASISs and midpoint of left and right posterior superior iliac spines (LPSIS and RPSIS) (or S1 joint of sacrum which is assumed to be the midpoint of PSISs) and it is positive in the forward direction.
- y-axis is proximally positive and orthogonal to x-z plane.

- **Thigh frame:**
  - Origin is the midpoint of lateral and medial epicondyles.
  - y-axis is positive in proximal and on the line joining the origin with the center of femoral head.
  - Quasi frontal plane is defined by y-axis and the epicondyles, and z-axis lies in this plane from left to right.
  - x-axis is orthogonal to y-z plane.

- **Shank frame:**
  - Origin is the midpoint of medial and lateral malleoli.
  - Quasi frontal plane is defined by the malleoli and head of fibula. The quasi sagittal plane is defined by the origin and tibial tuberosity and orthogonal to quasi frontal plane. y-axis is in the intersection of these two planes and oriented positively in proximal direction.
  - z-axis lies in the quasi frontal plane and positive from left to right.
  - x-axis is orthogonal to y-z plane and positive in forward direction.

- **Foot frame:**
  - Origin is calcaneus landmark.
  - y-axis is in the intersection of quasi transverse and quasi sagittal planes. Quasi transverse plane is defined by the origin, first and fifth metatarsal heads. Orthogonal to this and including origin and second metatarsal head is the quasi sagittal plane. y-axis is positive in proximal.
  - z-axis is positive from left to right and lies in the quasi transverse plane.
  - x-axis axis is orthogonal to y-z plane.
Another definition is proposed by International Society of Biomechanics in [29]. Reference frames for pelvis and thigh are nearly the same as those of Cappozzo et al. [22]. The only difference is the definition of origins. Wu et al. [29] defines both segments origins at the hip joint center. If joint translations are ignored, this difference does not give rise to any difference in segment orientations. Although these two publications define reference frames based on anatomical external landmarks, Conti et al. [26] showed the differences between the shank anatomic reference frame of both publications in vitro analysis. The definition of shank reference frame by Wu et al. is as follows:

- Origin is the midpoint of medial and lateral malleoli (LM and MM) the same as [22].
- z-axis is from IM to LM or MM depending on right or left leg.
- x-axis is perpendicular to the plane formed by origin and malleoli (LM and MM) and it is oriented anteriorly.
- y-axis is perpendicular to x-z plane

As stated by Conti et al. [26], the frontal plane of Cappozzo et al. [22] is formed by head of fibula and malleoli (LM and MM) while that of Wu et al. [29] contains midpoint of malleoli and epicondyles (MC and LC). The difference in the sagittal plane is that former uses tibial tuberosity while the latter uses midpoint of epicondyles both together with midpoint of malleoli. It turns out that these two reference frames have positive and negative aspects considering the repeatability. The fixed orientational differences between these two definitions should be noted when applied.

### 2.2.3 Joint Center Definitions

The joint centers of the hip, knee and ankle are important since they are used as the anatomical landmarks inside the body, thus called internal landmarks. Their positions in the global frame should be expressed to define the segmental anatomical frames. Joint centers are determined either by functional or predictive approaches.
Functional approaches are based on the movement data to find the joint centers. Cap-ppozzo in [23] stated that hip joint center is calculated by a test in which the subject moves his/her thigh through abduction-adduction and then flexion-extension. Using thigh marker positions in this movement hip joint center is determined by employing least-squares method. This approach assumes the hip joint as a ball-and-socket type joint. Joint center calculation is based on fitting the movement of markers on a sphere centered at the joint [30]. There are other functional approaches that make use of transformation matrices. In these methods, the residue of actual joint center and the joint center calculated from the marker positions by transforming to the appropriate reference frame (either global or pelvis depending on the assumption of pelvis movement) is minimized [31]. Piazza et al. [32] showed that the functional approach yields better results if the range of motion of hip joint is large. Predictive approaches, on the other hand, are based on the regression equations obtained from the radiographic ([18], [19]) or cadaveric studies [33]. An example of hip joint center estimation method is presented by Davis et al. [18]. The x-y-z coordinates of the hip joint center in the pelvis coordinate frame is

\[
x_{\text{hip}} = (-x_{\text{dis}} - r_{\text{marker}}) \cos \beta + C \cos \theta \sin \beta \tag{2.1}
\]

\[
y_{\text{hip}} = \sigma(C \sin \theta - \frac{d_{\text{asis}}}{2}) \tag{2.2}
\]

\[
z_{\text{hip}} = (-x_{\text{dis}} - r_{\text{marker}}) \sin \beta - C \cos \theta \sin \beta \tag{2.3}
\]

where \(d_{\text{asis}}\) is the distance between right and left ASIS, \(x_{\text{dis}}\) is anterior/posterior component of the ASIS/hip center distance and \(C = 0.115L_{\text{leg}} - 0.0153\). \(d_{\text{asis}}, x_{\text{dis}}\) and \(L_{\text{leg}}\) are measured quantities. \(r_{\text{marker}}\) is the marker radius. \(\sigma\) is +1 and −1 for the right and left hip, respectively. \(\theta\) and \(\beta\) are experimentally determined constants of 28.4° and 18°, respectively. There are regression equations for the knee and ankle joints in [18]. Vaughan et al. [19] have also described the regression equations to be used with anatomical frame definitions of the segments involved for the joint. Sandau et al. [34] revised these equations by comparing the results with MRI scans of the subjects. Harrington et al. [35] proposed new equations to be used in the deter-
mination of the hip joint center of patients with cerebral palsy. Anatomical frame
definition of the pelvis is the same as in [29]. The x, y and z components of the right
hip joint center is given as

\[ x = -0.24PD - 9.9 \]  \hspace{2cm} (2.4)\\
\[ y = -0.16PW - 0.04L_{leg} - 7.1 \]  \hspace{2cm} (2.5)\\
\[ z = 0.28PD + 0.16PW + 7.9 \]  \hspace{2cm} (2.6)

where PW is the same as \( d_{asis} \) and PD is the pelvic depth and defined as the distance
between the midpoints of the lines connecting the two ASIS and two PSIS. The latter
may be taken as the S1 joint of the sacrum.

New studies ([36] and [37]) suggest that if the hip range of motion is sufficiently
large, functional approach should be utilized to obtain more reliable results. If the
patient is not able to move the hip sufficiently, then the predictive approach should be
preferred.

2.2.4 Inverse Kinematic Description

The inverse kinematic phase of the analysis is performed to obtain segment orien-
tations with respect to global frame and joint angles. Relative orientation between
the segments that are connected by the joints is related to the joint angles. There are
various methods in the literature. Most commonly encountered method to express
segment rotations (either relative or absolute) is the Euler/Cardan angles due to its
clinical interpretation. This method is a well-known inverse kinematic description in
3-D kinematics and dynamics. Euler/Cardan angles make use of transformation ma-
trices to extract the angles. The transformation matrix is defined for a segment with
respect to the laboratory fixed frame. This is done by expressing its unit vectors in
terms of the unit vectors of the fixed frame.

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Figure 2.2 presents two reference frames $\mathcal{F}_A$ and $\mathcal{F}_B$ defined at points A and B, respectively. If $\mathcal{F}_B$ represents the segment and $\mathcal{F}_A$ is the fixed frame, then the transformation matrix becomes as in Equation 2.7 with the notation proposed by Özgören [38]. This notation is used throughout the thesis.

$$
\hat{C}^{(a,b)} = \hat{C} = \begin{bmatrix}
\hat{u}_1^{b/a} & \hat{u}_2^{b/a} & \hat{u}_3^{b/a}
\end{bmatrix} = \\
\begin{bmatrix}
c_{11} & c_{12} & c_{13} \\
c_{21} & c_{22} & c_{23} \\
c_{31} & c_{32} & c_{33}
\end{bmatrix}
$$

(2.7)

where $\hat{u}_n^{(b/a)}$ is defined as the $n^{th}$ unit vector of $\mathcal{F}_B$ expressed in $\mathcal{F}_A$. Joint angle is defined by the transformation matrix between proximal and distal segments. The proximal segment is considered as fixed and the distal segment moves relative to it through the joint. Thus, $\mathcal{F}_B$ is the proximal segment and $\mathcal{F}_A$ is the distal segment for the joint considered. After expressing the transformation matrix, an Euler/Cardan sequence is assumed to find distal segment’s orientation with respect to the proximal segment frame. If the rotations occur about the reoriented axes, then the sequence is called rotated frame based. The transformation matrix in Eq. 2.8 is defined as the first rotation about the first axis followed by the second rotation about the reoriented second axis and the third one about the twice reoriented third axis. 1, 2 and 3 in the expression may be interpreted as x, y and z if the axes are named by alphabet. In the expression, s and c refer to the sine and cosine of an angle, respectively.
\[ \dot{C} = \hat{R}_1(\theta_1)\hat{R}_2(\theta_2)\hat{R}_3(\theta_3) = e^{\hat{u}_1\theta_1}e^{\hat{u}_2\theta_2}e^{\hat{u}_3\theta_3} \]

\[
\begin{bmatrix}
    c\theta_2 c\theta_3 & -c\theta_2 s\theta_3 & s\theta_2 \\
    s\theta_1 s\theta_2 c\theta_3 + c\theta_1 s\theta_3 & -s\theta_1 s\theta_2 s\theta_3 + c\theta_1 c\theta_3 & -s\theta_1 c\theta_2 \\
    -c\theta_1 s\theta_2 c\theta_3 + s\theta_1 s\theta_3 & c\theta_1 s\theta_2 s\theta_3 & c\theta_1 c\theta_2
\end{bmatrix} \quad (2.8)
\]

By element matching, the angles can be calculated from the known rotation matrix. The following angles are obtained with double argument arctangent function defined as \( \text{atan2}(\sin \theta, \cos \theta) \) \(^1\)

\[
\theta_2 = \text{atan2}(c_{13}, \sigma \sqrt{1 - c_{13}^2}) \quad (2.9)
\]

\[
\theta_1 = \text{atan2}(-\sigma c_{23}, \sigma c_{33}) \quad (2.10)
\]

\[
\theta_3 = \text{atan2}(-\sigma c_{12}, \sigma c_{11}) \quad (2.11)
\]

There are two solutions of Equation \(2.9\) due to \( \sigma = \pm 1 \). If three angles are obtained as \(\theta_2\) and \(\theta_{1,3}\) when \(\sigma = +1\), the corresponding angles when \(\sigma = -1\) become \(\theta'_2 = \pi - \theta_2\) and \(\theta'_{1,3} = \pi + \theta_{1,3}\). The equations \(2.10\) and \(2.11\) for \(\theta_1\) and \(\theta_3\) involve the condition of \(\cos \theta_2\) being non-zero. If \(\cos \theta_2 = 0\) holds, then the so-called Gimbal Lock phenomenon occur. This is also called singular position. In this case, \(\theta_1\) and \(\theta_3\) cannot be obtained independently. By properly selected sequence of rotation for the analysis, this problem may be avoided for the range of motion (ROM) of the joint under consideration.

Most of the literature in gait analysis uses definition of Euler/Cardan angles interchangeably. However, MacWilliams et al. \([39]\) noticed this misunderstanding, and they argued that if the all rotations take place about different axes, that is, either one of x, y and z, such a sequence is called Cardan. On the other hand, in Euler sequence the first and last rotations are about the same (rotated or original) axis. As an example, rotation sequence of XYZ is named as Cardan while XYX is called Euler.

\(^1\) The double argument arctangent function is defined in terms of Matlab notation.
clarification is consistent with the original sequence of 3-1-3, i.e. ZXZ proposed by the 19th century Swiss mathematician Euler.

One of the earliest works using Cardan angles is by Grood et al. [40]. It is called “Grood and Suntay Joint Coordinate System” (JCS) in the literature. JCS is a coordinate frame that is made up of one axis from each distal and proximal segments reference frames and an axis, called floating, perpendicular to the other two. After defining segmental frames, they have defined joint rotations for the knee joint as flexion-extension occurring about femoral axis, internal-external rotation occurring about tibial axis and abduction-adduction occurring about the floating axis. They claimed that their definition of JCS is free from the sequence dependency of Euler/Cardan angles. Euler/Cardan angles may result in different numerical angle values for different sequences of the same angular orientation since finite angular rotations are not vectors (although infinitesimal ones are vectors). In fact, MacWilliams et al. have showed that JCS is a pre-defined sequence of Cardan. The sequence is in rotated frame based XYZ [39]. The rotation matrices obtained in JCS and Cardan sequence of XYZ result in the same joint angles except opposite signs in flexion-extension and abduction-adduction. The internal-external rotation of knee differs by 90°. Therefore, JCS also is sequence dependent as opposed to the claim in [40]. Due to its clinical interpretation and association with the clinical terminology of joint angles, this method has become widely accepted method of joint angle calculations ([15], [41]). The transformation matrix in JCS is defined in Eq. 2.12.

\[
\hat{C}_{JCS} = \begin{bmatrix}
  c\beta c\gamma & c\alpha s\gamma + s\alpha s\beta c\gamma & s\alpha s\gamma - c\alpha s\beta c\gamma \\
  -c\beta s\gamma & c\alpha c\gamma - s\alpha s\beta s\gamma & s\alpha c\gamma + c\alpha s\beta s\gamma \\
  s\beta & -s\alpha c\beta & c\alpha c\beta
\end{bmatrix}
\] (2.12)

Kadaba et al. [17] proposed a different transformation matrix to be used in joint angle calculations. Although their sequence is the same with that of JCS, the transformation matrix is different. Segmental axes defined in [17] and [40] differ by a constant rotation about the z-axis. That is in [40], lateral direction is x, anterior is y while z is the superior. On the other hand, in [17], y is laterally and x is anteriorly while superior direction is z. Eq. 2.13 explains the relationship between the transformation
matrices.

\[
\hat{C}_{Kadaba} = \hat{C}_{JCS} \hat{R}_z\left(-\frac{\pi}{2}\right)
\]  

(2.13)

The anatomical directions discussed above are presented in Figure 2.3 with the anatomical planes for the anatomical reference position. The right and left directions are lateral directions.

Figure 2.3: Anatomical Directions and Planes

If this manipulation in Eq. 2.13, the matrix obtained for [17] is given in Eq. 2.14.

\[
\hat{C}_{Kadaba} = \begin{bmatrix}
c\alpha s\gamma - s\alpha s\beta c\gamma & c\beta c\gamma & s\alpha s\gamma + c\alpha s\beta c\gamma \\
-c\alpha c\gamma - s\alpha s\beta s\gamma & c\beta s\gamma & -s\alpha c\gamma + c\alpha s\beta s\gamma \\
-s\alpha c\beta & -s\beta & c\alpha c\beta
\end{bmatrix}
\]  

(2.14)

The transformation matrices in Equations 2.12 and 2.14 become identical if the angle definitions are revised as follows:

- **Flexion** is the same as \(\alpha_{JCS} = \alpha_{Kadaba}\)
- **Abduction** is \(\beta_{JCS} = -\beta_{Kadaba}\)
- **Rotation** is \(\gamma_{JCS} = \gamma_{Kadaba} - \frac{\pi}{2}\)
The same sequence is applied in \[42\] by Tupling et al. using a transformation matrix different than the others. As discussed above for Eq. 2.14 the same applies here. The necessary transformation between reference frames is done by a half rotation about y and x axes. The transformation matrix Eq. 2.12 and that in \[42\] compared and followings are noted:

- Flexion is $\alpha_{\text{JCS}} = -\alpha_{\text{Tupling}}$
- Abduction is $\beta_{\text{JCS}} = \frac{\pi}{2} + \beta_{\text{Tupling}}$
- Rotation is $\gamma_{\text{JCS}} = \frac{\pi}{2} + \gamma_{\text{Tupling}}$

Cappozzo et al. \[25\] argued that the applied sequence is the same as JCS in their work. Since the definition of segmental axes were different, the transformation matrices obtained would be different. The constant transformations between the segmental reference frame of \[40\] and \[25\] take place around x-axis and around z-axis both by 90°. Thus obtained transformation matrix and the transformation matrix obtained by Cappozzo et al. \[25\] differs by the following definitions of joint angles:

- Flexion and abduction have changed the roles as:
  $\alpha_{\text{JCS}} = \beta_{\text{Cappozzo}}$
  and
  $\beta_{\text{JCS}} = \alpha_{\text{Cappozzo}}$
- Rotation becomes $\gamma_{\text{JCS}} = \beta_{\text{Cappozzo}} - \frac{\pi}{2}$

Another inverse kinematic solution is to apply Denavit – Hartenberg (DH) convention used in robotics. Apkarian et al. \[43\] used DH convention by assuming there are two internal links inside a spherical joint. Joint model is composed of four links including distal and proximal segments. Each link included in a joint is connected by a revolute joint. Therefore, the total degree of freedom of a joint is three. The segmental reference frames are the same as in \[25\]. Sequence of rotations is defined as abduction-adduction followed by internal-external rotation and flexion-extension. JCS and \[43\] yield the identical results by defining following
Abduction and internal rotation are switched as:

\[ \beta_{JCS} = \gamma_{Apkarian} - \frac{\pi}{2} \]

and

\[ \gamma_{JCS} = \beta_{Apkarian} - \pi \]

Flexion is changed by extension as \( \alpha_{JCS} = -\alpha_{Apkarian} \)

Euler/Cardan angle solutions are dependent on the rotation sequence. As the sequence changes, the results will change. Most of the literature uses the flexion-abduction-internal rotation sequence.

Another solution method is the use of helical screw method [44]. This method finds an axis about which all the joint rotations occur as a single rotation. If the translation is allowed, the screw axis is the axis along which the translations occur as a single translation. By applying the Rodrigues formula in robotics, the unit vector of the axis and the amount of rotation is calculated. Given the transformation matrix, the amount of rotation \( \phi \) is found by Equations 2.15 and 2.16

\[
\cos \phi = \frac{Tr(\hat{C}) - 1}{2} \quad (2.15)
\]

\[
\sin \phi = \frac{1}{2} \sqrt{(c_{21} - c_{12})^2 + (c_{32} - c_{23})^2 + (c_{31} - c_{13})^2} \quad (2.16)
\]

The elements of the unit vector of joint axis are obtained as in Equation set 2.17

\[
\begin{align*}
n_1 &= \sigma \sqrt{\frac{c_{11} - \cos \phi}{1 - \cos \phi}} \\
n_2 &= \sigma \sqrt{\frac{c_{22} - \cos \phi}{1 - \cos \phi}} \\
n_3 &= \sigma \sqrt{\frac{c_{33} - \cos \phi}{1 - \cos \phi}}
\end{align*} \quad (2.17)
\]

Due to wide acceptance on the Euler/Cardan solutions of joint angles, it is the preferred solution despite the differences in joint angles with different sequences employed.
2.2.5 Inverse Dynamics

Inverse dynamics is the solution of the question “given the movement, what are the forces and moments required to produce and accompany the motion?”. The forces asked for are the internal joint reaction forces ($\vec{F}$) and moments ($M_C$). The Newton-Euler method is applied to compute them with the information of ground reaction forces. Equations 2.18 and 2.19 are written for each segment starting from foot. Similar to the kinematics, dynamics or kinetic analysis is based on some assumptions and idealizations. First of all, bodies are considered to be rigid with a constant mass moment of inertia relative to body fixed coordinates. Each joint is idealized to be spherical type joint to obtain single point of force application at the joint. The friction in the joints and shock absorption characteristics of cartilage are ignored for computational purposes [15].

\[
m\ddot{a}_C = \Sigma \vec{F} \quad (2.18)
\]

\[
\vec{J}_C \cdot \dot{\alpha} + \vec{\omega} \times \vec{J}_C \cdot \dot{\vec{ω}} = \Sigma \vec{M}_C \quad (2.19)
\]

where C is the center of mass of the segment under consideration. The segmental angular velocities ($\vec{ω}$) and both linear acceleration of centers of mass and angular accelerations ($\vec{a}_C$ and $\vec{α}$) are to be found from previous kinematic analysis. Angular velocity is defined in Eq. 2.20

\[
\vec{ω} = \hat{C} \hat{C}^T \quad (2.20)
\]

where skew-symmetric angular velocity matrix is defined from its column vector in the form $\vec{ω} = \begin{bmatrix} \omega_1 & \omega_2 & \omega_3 \end{bmatrix}^T$ in Eq. 2.21

\[
\vec{ω} = \begin{bmatrix} 0 & -\omega_3 & \omega_2 \\ \omega_3 & 0 & -\omega_1 \\ -\omega_2 & \omega_1 & 0 \end{bmatrix} \quad (2.21)
\]
Since the transformation matrix is defined numerically for each frame, its time derivative can be obtained by numerical differentiation. The angular acceleration is found through a numerical differentiation of angular velocity components.

\[
\ddot{\alpha} = \begin{bmatrix} \alpha_1 & \alpha_2 & \alpha_3 \end{bmatrix}^T = \begin{bmatrix} \dot{\omega}_1 & \dot{\omega}_2 & \dot{\omega}_3 \end{bmatrix}^T
\] (2.22)

The locations of center of mass of the segments are required. In literature, the equation presented in Eq. 2.23 is used to find the center of mass location, where \( \bar{p}_{\text{proximal}} \) and \( \bar{p}_{\text{proximal}} \) are the proximal and distal joint center coordinates and \( c \) is the coefficient obtained from cadaveric studies using regression equations. Different coefficients are proposed in different studies \[19], \[45] and \[46].

\[
\bar{p}_C = \bar{p}_{\text{proximal}} + c(\bar{p}_{\text{distal}} - \bar{p}_{\text{proximal}})
\] (2.23)

The constant anthropometric properties mass (m), mass moment of inertia tensor (\( \overline{I_C} \)) are also required to solve the associated equations. These are found on the basis of cadaveric studies or through idealized geometries and mathematical modeling. The masses of the segments are taken as a percentage of total body mass \[43], \[45]. Apkarian et al. \[43] used a single percentage, while de Leva \[45] used different percentages for female and male subjects. For the mass moment of inertia, Apkarian et al. \[43] suggested idealized mathematical modelling. The thigh and shank segments are modelled as conic sections while the foot is modelled as a cylinder. De Leva \[45], on the other hand, gave the radius of gyration of segments in each direction as a percentage of segment lengths. Later, Futamure et al. \[47] showed that joint moments can be calculated without using inertia parameters due to the fact that its effect is at most 10%. Similar observations are made by Erer \[8], who showed that for a normal human gait at a normal speed, the inertia effects are negligible in the stance phase when compared to the ground reaction forces.

Lastly, the inputs to the Newton-Euler equations are the ground reaction forces and moments. There are some methods developed to compute ground reaction forces without using any device to measure the quantities. In the swing phase of gait where there is only one foot in contact with the ground, the equation is obtained by applying
the Newton’s Second Law to the whole body,

\[ \vec{F}_{gr} + \vec{F}_{gl} = \sum_{i=1}^{n} m_i (\vec{a}_i - \vec{g}) \]  

(2.24)

where \( \vec{F}_{gr} \) and \( \vec{F}_{gl} \) are the ground reaction forces applied on the right and left foot, respectively, and \( i \) denotes the number of segments whose masses and mass center accelerations are denoted by \( m_i \) and \( \vec{a}_i \), respectively. In the swing phase, the swinging foot is considered to have little effect on the ground reaction force due to relatively small magnitude of inertia forces, and ground reaction forces and moments can readily be calculated by using the Euler form of the equation. However, in double stance, the equation becomes indeterminate. The solution of this problem is described by Ren et al. [48]. The method of “Smooth Transition Assumption” (STA) is proposed considering the whole body. With this approach, they have achieved good results on the sagittal plane forces and moments. Ground reaction forces and moments in double stance phase is empirically estimated by exponential functions except the force in the walking direction since it is not monotonic. In the swing phase, ground reaction forces on the foot in contact with the ground are calculated from Equation 2.24. When the other foot makes contact with the ground, the STA is utilized. By the information of these forces, forces on the previously contacting foot is again calculated by the equation above. The same idea applies for the moments.

The force plates are used extensively in the gait analysis laboratories since their accuracy is superior compared to that of the motion capture system, except there exists the distraction of subject targeting to step on the plate [15]. However, in case of non-laboratory environments, the STA methods may be well suited together with the good estimations of segmental inertia parameters.

2.3 Instrumental and Experimental Errors

Instrumental errors are due to photogrammetric measurements in the case of camera based marker systems. In other words, the construction of 3-D position of a marker from 2-D images involves error. These errors are classified in two groups by Chiari et al. [49]. The systematic error is related to the camera calibration model and accuracy of the calibration parameter estimation. The other one is called random error due
to electronic noise and conversion of marker image coordinates into marker position coordinates. These errors can be eliminated by either the carefully designed setup in the case of systematic error or smoothing the data in the case of random error. Up to 1 mm resolution may be achieved by a carefully designed setup.

Experimental errors are caused by the experiment itself. One of the most important error sources is the skin movement artefact which may reach up to 40 mm for lateral epicondyle marker in the antero-posterior direction. These may lead to misleading joint rotations due to erroneous orientation of segments. In external-internal rotation angles, the skin motion artefact may cause an error equal to the range of motion. The photogrammetric artefacts are less compared to that of skin movement. Cappozzo et al. [22] emphasized that filtering is not a solution to this error because the errors have the same frequency content as that of the underlying bone.

2.4 Simulation of Gait

The gathered data through one of the gait analysis protocols is processed with a software developed based on a biomechanical model. The model here is referred to a musculoskeletal model including bones and muscles. Some commercial software packages like Matlab-Simulink and Adams Life Modeller provide such biomechanical models. These are mainly developed for general purposes and implemented to gait analysis as well. However, in 2007, Stanford University presented an open-source software called OpenSim to analyze and simulate gait of a model which is mainly a person [50]. The marker data together with the ground reaction forces and moments constitute input to OpenSim, which then conducts the inverse kinematic analysis together with the inverse dynamic analysis. The muscle forces or resulting moments during walking can also be calculated. After the regular analysis for a subject, the main capability of the software comes into play. It is the forward dynamic analysis which may include “what-if” type scenarios. The forward dynamics may be a check for the previous inverse analysis. “What-if” type scenarios may lead to complex studies such as muscle-lengthening surgery planning or the improvement of walking characteristics under the alterations on a joint. The implementation of regular gait analysis data to OpenSim and its main features are discussed in the preceding
2.4.1 Data Types and Importing

The first input is the marker data in trc (abbreviated from Track-Row-Column) file format which resembles Excel file format, given in Figure 2.4.

![Figure 2.4: Marker Data File for OpenSim](image)

The first three rows are header including the name of the subject, and data acquisition properties. Next two rows followed by marker data include the frame numbers and time together with the name of marker data. The second input is the ground reaction forces and moments given in a motion file (abbreviated as mot) which resembles trc file in text format. These two inputs are necessary and sufficient data to run a simulation on OpenSim. However, some other data may be used to improve and check the simulation steps and results. The additional data may be EMG signals and joint angles. EMG data may be included only for some of the muscles. Joint
angles may be calculated in another software by using a certain biomechanical model. Since joint angle calculation process of OpenSim relies on least square method, the weightings of the marker data can be tuned according to the joint angles calculated outside OpenSim.

### 2.4.2 Scaling of The Generic Model

The generic model of OpenSim is created for a specific subject which includes the segmental bones and muscles. There are different types of generic models depending on the segments and muscles involved, such as upper and lower extremity and whole body as well. These generic models are free from the marker set and biomechanical model. The generic models are scaled for the subject under consideration using the static shot marker data. Static shot is not necessarily a single source of data. It may be taken during a time interval and then averaged to use in scaling. There are two types of markers in OpenSim. One of them is the model marker which is related to the generic model and its segmental properties like its length. The other marker type is called experimental marker which is the input data, obtained, for example, from camera images. Yet, another type is called virtual marker. Virtual marker is a subset of experimental markers. It may include joint centers and the projections of some markers on to a plane.

### 2.4.3 Inverse Kinematics

Inverse kinematics is solved by minimizing the weighted square of difference between experimental and model marker locations subject to joint constraints \[\text{[51]}\] as presented in Equation 2.25. The joint angles may also be included in this step. The multipliers \(w_i\) and \(w_j\) are used to weigh the markers and angles differently.

\[
e = \sum_{i=1}^{\text{markers}} w_i (\vec{x}_{i}^{\text{exp}} - \vec{x}_{i}^{\text{model}})^2 + \sum_{j=1}^{\text{jointangles}} \omega_j (\theta_{j}^{\text{calculated}} - \theta_{j}^{\text{model}})^2 \tag{2.25}
\]

Experimentally measured marker positions in 3-D and calculated joint angles are denoted by \(\vec{x}_{i}^{\text{exp}}\) and \(\theta_{j}^{\text{calculated}}\) for the \(i\)th marker and \(j\)th joint angle, respectively. Those
associated with the model are $\vec{x}_{\text{model}}^i$ and $\vec{\theta}_{\text{model}}^j$. The weightings are denoted by $w_i$ and $\omega_j$. Marker weighting, $w_i$, is a vector with three elements whose dot product is taken with the marker error square to obtain a single error value corresponding to the $i^{th}$ marker. The joint angle weighting, $\omega_j$, is a scalar value. The joint angles producing the minimum error defined above is obtained. Since the whole body is under consideration, Lu et al. [52] called this as Global Optimization Method (GOM).

The aforementioned conventional gait model is referred as direct method (DM). DM does not involve joint constraint violations due to skin movement artefact. An improvement on this method is the work of Cappello et al. [53] and Challis [54]. They applied a segmental optimization approach (SOM).

$$\min f = \sum_{i=1}^{m} (\hat{R}x_i + v - y_i)^T(\hat{R}x_i + v - y_i) \quad (2.26)$$

The rotation matrix $R$ is obtained via optimization. $x_i$ and $y_i$ are the position vectors of $i^{th}$ marker at the current and the global reference frame and $v$ is the translation vector. It is minimized during a translation and rotation of rigid body.

$$y_i = \hat{R}x_i + v \quad (2.27)$$

SOM improves the DM by taking the skin motion artifact into account. However, it does not include joint constraints. Although the segments are defined by using the common joint centers, this constraint is not truly matched due to skin movement artefact and experimental errors. SOM only cares about the segmental skin motion errors.

GOM proposed by Lu et al., [52] solves the least squares problem throughout the body. In a standing upright position, subject’s markers are taken as the reference to customize the subject-specific model. Generalized coordinates are defined as the translational or rotational variables of joint depending on its type. $P$ contains the measured marker coordinates, $P^\ast$ is the marker coordinates in segmental frames and $W$ is the weighting.
Minimization of the above function will yield the joint angles. Segmental weighting factor can be calculated as 1 divided by its residual error. Since the skin artefact is higher for the thigh with larger residual error [22], the weighting factor associated with it will be less than the others. Above minimization is done numerically which needs initial guesses. It may be obtained from SOM approach. Lu et al. [52] have designed three links representing pelvis, thigh and shank apparatus with two spherical type joint representing the hip and knee joint centers. The joint angles are applied to the model, and the markers’ positions are obtained. Noise is introduced to the data obtained by different type of skin movement artefact artificially. GOM reproduces the actual motion better than other two approaches.
CHAPTER 3

KISS-GAIT ANALYSIS SYSTEM OF METU

The gait analysis system of METU, called KISS, is the first gait analysis system of Turkey. The KISS system will be explained in three parts as experimental procedure, inverse kinematics solution of joint angles and definition of foot angles. The last section of this chapter investigates effects of different sequences in joint angle calculations.

3.1 Experimental Procedure

KISS has six monochrome cameras with wide-angle lenses and interlaced PAL resolution (576i) for positional measurements which is used in inverse kinematics calculations and two force plates to measure ground reaction forces which is used in the inverse dynamics calculations. The marker set is Helen-Hayes marker set. The markers are captured by six cameras equipped with infrared light emitting diodes (LEDs) having an adjustable duty-cycle running at 50 Hz. Retroreflective material is used to coat the markers (balls with 12.7 mm, or half inch, radius). Just markers are captured by the filter that passes only infrared light reflected. Since the cameras are not fixed to the ground, but on tripods, the calibration procedure is needed before every experiment. By knowing the positions of the markers on the calibration rods (in a fixed position in the calibration volume) as shown in Figure 3.1, the constant parameters which will be used in the regeneration of 3-D positions of points seen in 2-D images are calculated. Also, a linearization grid is used to eliminate distortions in the 2-D images due to wide angle camera lenses (Figure 3.1). Anthropometric measurements are the last step before a gait trial.
The following measurements are taken on the subject before every test:

- **DASIS**: Distance from right anterior superior iliac spine (RASIS) to left anterior superior iliac spines (LASIS)
- **Leg Length**: Distance from RASIS (or LASIS) to right (or left) medial malleoli (MM) on the line passing through femoral epicondyle (FE) on the medial side
- **Knee Width**: Distance from medial FE to lateral FE of the right and the left knee
- **Ankle Width**: Distance from medial malleoli (MM) to lateral malleoli (LM)
- **Other parameters**: Mass and height of the subject

The experimental procedure has two main parts. In the static shot, subject stands still in the upright position. The markers used in the static shot are presented in Table 3.1 which is somewhat different than the second phase called dynamic trial, in which subjects walks and markers are captured together with the ground reaction forces. The markers of dynamic trial are also given in Table 3.1. Dynamic trials are repeated until the subject correctly steps on both of the force plates. A subject must step on each of the force plates separately. Also, the foot should be fully within the top plate of the force plate so that there is no force flow to the ground and ground reaction force components are measured by the force plate correctly.
Table 3.1: Markers Used in Static (s) and Dynamic (d) Trials

<table>
<thead>
<tr>
<th>Marker Name</th>
<th>Marker Position</th>
<th>Trial</th>
</tr>
</thead>
<tbody>
<tr>
<td>RASIS</td>
<td>Right Anterior Superior Iliac Spine</td>
<td>s,d</td>
</tr>
<tr>
<td>LASIS</td>
<td>Left Anterior Superior Iliac Spine</td>
<td>s,d</td>
</tr>
<tr>
<td>SACRUM</td>
<td>Midpoint between Posterior Iliac Spines</td>
<td>s,d</td>
</tr>
<tr>
<td>RTHIGH</td>
<td>Right Thigh Wand</td>
<td>s,d</td>
</tr>
<tr>
<td>LTHIGH</td>
<td>Left Thigh Wand</td>
<td>s,d</td>
</tr>
<tr>
<td>RKNEE</td>
<td>Right Lateral Femoral Epicondyle</td>
<td>d</td>
</tr>
<tr>
<td>LKNEE</td>
<td>Left Lateral Femoral Epicondyle</td>
<td>d</td>
</tr>
<tr>
<td>ROKCD</td>
<td>Right Outer Marker of Knee Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>RIKCD</td>
<td>Right Inner Marker of Knee Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>LOKCD</td>
<td>Left Outer Marker of Knee Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>LIKCD</td>
<td>Left Inner Marker of Knee Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>RSHANK</td>
<td>Right Shank Wand</td>
<td>s,d</td>
</tr>
<tr>
<td>LSHANK</td>
<td>Left Shank Wand</td>
<td>s,d</td>
</tr>
<tr>
<td>RANKLE</td>
<td>Right Lateral Malleolus</td>
<td>d</td>
</tr>
<tr>
<td>LANKLE</td>
<td>Left Lateral Malleolus</td>
<td>d</td>
</tr>
<tr>
<td>ROACD</td>
<td>Right Outer Marker of Ankle Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>RIACD</td>
<td>Right Inner Marker of Ankle Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>LOACD</td>
<td>Left Outer Marker of Ankle Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>LIACD</td>
<td>Left Inner Marker of Ankle Centering Device</td>
<td>s</td>
</tr>
<tr>
<td>RHEEL</td>
<td>Right Heel</td>
<td>s,d</td>
</tr>
<tr>
<td>LHEEL</td>
<td>Left Heel</td>
<td>s,d</td>
</tr>
<tr>
<td>RMETA2</td>
<td>Right Second Metatarsal</td>
<td>s,d</td>
</tr>
<tr>
<td>LMETA2</td>
<td>Left Second Metatarsal</td>
<td>s,d</td>
</tr>
</tbody>
</table>
The reference frame with which the data is collected and global reference frame of the laboratory are different. The data collection takes place in the reference frame whose y-axis is opposite to the walking direction, x-axis is to the left of the subject when walking and z-axis is directed upwards as shown in Figure 3.2 by $X_D$ and $Y_D$. The laboratory fixed global frame is oriented such that its x-axis is in the walking direction, z-axis is to the right of the subject when walking and y-axis is upwards as shown in Figure 3.2 by $X_G$ and $Z_G$.

Therefore, collected data is transformed to the global frame by means of a simple rotation matrix denoted by $\hat{R}$. This transformation can be defined as a half rotation ($\pi/2$ rad) about the y-axis and then about the x-axis. The transformation matrix in Eq.
3.1 is obtained.

\[
\hat{C} = \hat{R}_y(\frac{\pi}{2}) \hat{R}_x(\frac{\pi}{2}) = \begin{bmatrix}
0 & -1 & 0 \\
0 & 0 & 1 \\
-1 & 0 & 0
\end{bmatrix}
\] (3.1)

Since the gait analysis protocol of KISS system makes use of Helen-Hayes marker set, the segmental anatomical frames are defined based on the technical reference frames that are defined by using the technical and then anatomical markers. The main assumption is that the orientation of the anatomical planes remains constant with respect to the orientation of the technical markers. Therefore, both anatomical and technical frames are defined in the static shot. Since relative orientation of anatomical frame with respect to technical frame can be calculated at this static shot, anatomical frames in the dynamic walking can be reconstructed from the known relationship between anatomical and technical frames. Let \( \hat{C}^{(A,i)} \) and \( \hat{C}^{(A,i^*)} \) denote anatomical to global and technical to global transformation matrices where \( i \) denotes the segment, the transformation matrix from technical to anatomical frames for the segment is calculated as in Eq. 3.2.

\[
\hat{C}^{(i^*,i)} = \hat{C}^{(A,i^*)T} \hat{C}^{(A,i)}
\] (3.2)

The technical frames in the dynamic walking experiment is pre-multiplied by this matrix to obtain the transformation matrix for the anatomical frame of each segment. Pelvis is defined by three anatomical landmarks during the static shot and dynamic walking. Therefore, technical and anatomical frames are the same for the pelvis and is defined as follows:

- Origin is the midpoint between right and left ASIS
- z-axis is from LASIS to RASIS
- y-axis is defined as the perpendicular to the plane defined by the three markers of the pelvis and it is directed superiorly
- x-axis is orthogonal to the plane defined by y and z axes in a right handed manner
Technical reference frame for the thigh segment is defined as follows,

- Origin is the knee marker
- y-axis is directed from knee marker to the hip joint center
- x-axis is in the anterior direction and perpendicular to the plane defined by thigh, knee markers and the hip joint center
- z-axis is orthogonal to the plane defined by y- and z-axes in a right handed manner

Anatomical frame for the thigh segment is defined as follows,

- Origin is the knee joint center
- y-axis is directed from knee joint center to the hip joint center
- x-axis is in the anterior direction and perpendicular to the plane defined by the hip and knee joint centers and the knee axis
- z-axis is orthogonal to the plane defined by y- and z-axes in a right handed manner

Technical reference frame for the shank segment is defined as follows,

- Origin is the ankle marker
- y-axis is directed from ankle marker to the knee joint center
- x-axis is in the anterior direction and perpendicular to the plane defined by shank, ankle markers and the knee joint center
- z-axis is orthogonal to the plane defined by y- and z-axes in a right handed manner

Anatomical frame for the shank segment is defined as follows,
• Origin is the ankle joint center
• y-axis is directed from ankle joint center to the knee joint center
• x-axis is in the anterior direction and perpendicular to the plane defined by the knee and ankle joint centers and the ankle axis
• z-axis is orthogonal to the plane defined by y- and z-axes in a right handed manner

3.2 Inverse Kinematics

The segments’ orientation and joint angles are obtained through Denavit-Hartenberg convention. In the joint angle calculation, it is assumed that there are two bodies between the proximal and distal segments of the joint. Thus, there are three revolute joints located between proximal and first intermediate body, first and second intermediate bodies, and second intermediate body and the distal body. Since the intermediate bodies are just imaginary bodies with zero link lengths, the assembly shown in Figure 3.3 is equivalent to a spherical joint between proximal and distal bodies.

The axis, $\overrightarrow{u_{3}}^{(1)}$, corresponds to the floating axis of the JCS discussed in Section 2.2.4. The rotation sequence employed by KISS is obtained in Eq. 3.3

$$\hat{C} = \hat{R}_z(\theta_1) \hat{R}_x(-\frac{\pi}{2}) \hat{R}_z(\theta_2) \hat{R}_x(-\frac{\pi}{2}) \hat{R}_z(\theta_3) \hat{R}_x(-\frac{\pi}{2})$$  (3.3)
whose matrix form is presented in Eq. 3.4

\[
\hat{C} = \begin{bmatrix}
    c\theta_1 c\theta_2 c\theta_3 + s\theta_1 s\theta_3 & c\theta_1 s\theta_2 & -c\theta_1 c\theta_2 s\theta_3 + s\theta_1 c\theta_3 \\
    -s\theta_1 c\theta_2 c\theta_3 - c\theta_1 s\theta_3 & s\theta_1 s\theta_2 & -s\theta_1 c\theta_2 s\theta_3 - c\theta_1 c\theta_3 \\
    s\theta_2 c\theta_3 & c\theta_2 & s\theta_2 s\theta_3
\end{bmatrix}
\] (3.4)

where \(\theta_1, \theta_2\) and \(\theta_3\) are the joint rotations about the medio-lateral, antero-posterior and infero-superior directions, respectively. For the hip joint, \(\theta_1, \theta_2\) and \(\theta_3\) denote flexion-extension, abduction-adduction and internal-external rotation, respectively. The pelvis angles are obtained as presented in Eq. set 3.5 where tilt and obliquity are around medio-lateral and antero-posterior axes, respectively,

\[
\begin{aligned}
    Tilt &= 90^\circ - \theta_1 \\
    Obliquity &= 90^\circ - \theta_2 \\
    Rotation &= 90^\circ - \theta_3
\end{aligned}
\] (3.5)

The hip joint angles are defined as follows in Eq. 3.6

\[
\begin{aligned}
    HipFlexion &= \theta_1 - 90^\circ \\
    HipAbduction &= \theta_2 - 90^\circ \\
    HipRotation &= 90^\circ - \theta_3
\end{aligned}
\] (3.6)

The knee joint angles are obtained from Eq. 3.7 where valgus is the knee abduction.

\[
\begin{aligned}
    KneeFlexion &= 90^\circ - \theta_1 \\
    KneeValgus &= \theta_2 - 90^\circ \\
    KneeRotation &= 90^\circ - \theta_3
\end{aligned}
\] (3.7)

It should be noted that the Denavit-Hartenberg convention described above is in fact equivalent to a specific Euler/Cardan angle sequence. If the rotation sequence is given as

\[
\tilde{S}_a \xrightarrow{\text{rot}[\vec{v}^{(n)}_x \beta_1]} \tilde{S}_m \xrightarrow{\text{rot}[\vec{v}^{(m)}_y \beta_2]} \tilde{S}_n \xrightarrow{\text{rot}[\vec{v}^{(n)}_z \beta_3]} \tilde{S}_b
\]
Then, the Euler/Cardan angle transformation matrix in Eq. (3.8) is defined in terms of exponential rotation matrices.

\[
\hat{C}^{(a,b)} = \hat{R}_i(\theta_i)\hat{R}_j(\theta_j)\hat{R}_k(\theta_k) = e^{\hat{u}_i\theta_i}e^{\hat{u}_j\theta_j}e^{\hat{u}_k\theta_k}
\]  

(3.8)

It will yield the same angles with the rotation sequence of ZXY since in the global frame it corresponds to the sequence of flexion-abduction-rotation which is the same sequence used in KISS-Gait. ZXY sequence is defined by the matrix presented in Eq. (3.9)

\[
\hat{C} = \begin{bmatrix}
-s\theta_1 s\theta_2 c\theta_3 + c\theta_1 s\theta_3 & -c\theta_1 s\theta_2 c\theta_3 + s\theta_1 s\theta_2 s\theta_3 & -s\theta_1 s\theta_2 c\theta_3 + c\theta_1 s\theta_2 s\theta_3 \\
 s\theta_1 s\theta_2 c\theta_3 + c\theta_1 c\theta_3 & c\theta_1 c\theta_3 - s\theta_1 c\theta_2 c\theta_3 + s\theta_2 s\theta_3 & -s\theta_1 c\theta_2 c\theta_3 + c\theta_1 s\theta_2 s\theta_3 \\
 -c\theta_1 s\theta_2 & s\theta_1 & c\theta_1 c\theta_2
\end{bmatrix}
\]  

(3.9)

If the transformation matrix for ZXY sequence is compared by the KISS transformation matrix, their equivalence will be revealed by Equations (3.10), (3.11) and (3.12), where \(\theta_{i,KISS}\) denote the KISS results.

\[
\theta_1 = 90 - \theta_{2,KISS}
\]  

(3.10)

\[
\theta_2 = 90 - \theta_{3,KISS}
\]  

(3.11)

\[
\theta_3 = 90 - \theta_{1,KISS}
\]  

(3.12)

Therefore, as expected from the sequence, flexion, abduction and rotation angles are \(\theta_3, \theta_1, \text{ and } \theta_2\), respectively.

### 3.3 Definition of Foot Angles

Different from other joint angle calculations, the foot angle calculation does not rely on Euler Angles. The angle between the vectors combining second metatarsal to the
heel and the vector combining second metatarsal to the ankle joint center is called static plantar flexion angle of the foot and is calculated at the static shot. In the dynamic session of the experiment, another angle is calculated as the angle between the vector combining second metatarsal to the ankle joint center and x-axis of shank anatomical frame. Foot dorsiflexion is defined as the sum of this angle and the static plantar flexion angle. The dynamic portion of dorsiflexion angle occurs nearly around the z-axis of global frame. Rotation of the foot around the y-axis of shank anatomical frame is defined as foot rotation, and rotation of the foot around y-axis of global frame of laboratory is defined as foot alignment angle.

3.4 Effect of Rotation Sequences on Inverse Kinematics

The idea to perform such an analysis is originated from the ambiguity in the literature. There are many researchers who have used different notations and angle sequences to obtain a clinically meaningful set of joint angles. First of all, it should be noted that the Euler/Cardan solutions are dependent on the sequence defined. That is, different sequence of rotations will result in different angles for the joint under consideration. This is illustrated in Figure 3.4, where two different sequences of rotations are considered for a two-body system. The rotation sequences shown result in different end poses of the bodies with respect to the original pose, although the angles $(50^\circ, 40^\circ$ and $30^\circ$ around x,y and z axes, respectively) are the same for both sequences.

Figure 3.4: Two Different Sequences of Rotations
Looking from another perspective, it may be observed that a body rotated in a sequence, say XYZ, through certain angles in each axis will have different orientation if the body is rotated backwards with another angle sequence but with the same amount of angles. The original pose seen in Figure 3.5 is rotated in XYZ sequence with certain angle values around certain axes to obtain the second pose. To obtain the original pose back, the rotation sequence YXZ is utilized. The angle values around the axes are kept constant. As expected, the third pose is not the same as the original pose due to different rotation sequences used.

![Figure 3.5: Different Poses Resulted From Different Rotation Sequences](image)

For the rotation from original pose to second pose seen in Figure 3.5, the rotation matrix, $\hat{C}_1$, is defined in Equation 3.13a. The rotation sequence is XYZ with the values of $50^\circ$, $40^\circ$, and $30^\circ$. If the pose is rotated backwards with ZYX sequence, with the angles $-50^\circ$, $-40^\circ$, and $-30^\circ$, the transformation matrix, $\hat{C}_2$, is as given in Equation 3.13b. With the same angles, but in the YXZ order, the transformation matrix, $\hat{C}_3$, becomes the one in Equation 3.13c.
\[
\hat{C}_1 = \hat{R}_z(50^\circ)\hat{R}_y(40^\circ)\hat{R}_z(30^\circ) = \begin{bmatrix}
0.663414 & -0.383022 & 0.642788 \\
0.747828 & 0.310468 & -0.586824 \\
0.025201 & 0.870002 & 0.492404
\end{bmatrix}
\] (3.13a)

\[
\hat{C}_2 = \hat{R}_z(-30^\circ)\hat{R}_y(-40^\circ)\hat{R}_z(-50^\circ) = \begin{bmatrix}
0.663414 & 0.747828 & 0.025201 \\
-0.383022 & 0.310468 & 0.870002 \\
0.642788 & -0.586824 & 0.492404
\end{bmatrix}
\] (3.13b)

\[
\hat{C}_3 = \hat{R}_y(-40^\circ)\hat{R}_z(-50^\circ)\hat{R}_z(-30^\circ) = \begin{bmatrix}
0.41721 & 0.80946 & -0.41318 \\
-0.32139 & 0.55667 & 0.76604 \\
0.85008 & -0.18681 & 0.49240
\end{bmatrix}
\] (3.13c)

The first backward rotations of ZXY will result in an orientation matrix of \(\hat{C}_{zxy}\) which is the multiplication of \(\hat{C}_1\) and \(\hat{C}_2\). On the other hand, \(\hat{C}_{yxz}\) orientation matrix is calculated by the multiplication of \(\hat{C}_1\) and \(\hat{C}_3\). The two orientation matrices are calculated as shown in Equations 3.14a and 3.14b. If \(\hat{C}_{zxy}\) was an identity matrix, the same initial pose would be obtained. However, \(\hat{C}_{yxz}\) is not an identity matrix, and therefore a pose different than the initial pose will be obtained by the backward rotations of the same amounts but with a different sequence.

\[
\hat{C}_{zxy} = \hat{C}_1\hat{C}_2 = \begin{bmatrix}
1 & 0 & 0 \\
0 & 1 & 0 \\
0 & 0 & 1
\end{bmatrix}
\] (3.14a)

\[
\hat{C}_{yxz} = \hat{C}_1\hat{C}_3 = \begin{bmatrix}
0.94631 & 0.20371 & -0.25101 \\
-0.28663 & 0.88779 & -0.36011 \\
0.14949 & 0.41272 & 0.89851
\end{bmatrix}
\] (3.14b)

Therefore, it is crucial to define the rotation sequence when interpreting the kinematic results of a gait trial.

In what follows, the effects of rotation sequences are investigated for a typical gait data obtained from KISS. The reference is taken as the ZXY sequence, which has been shown to be equivalent to the sequence used in KISS system.
The results are presented in Figures 3.6, 3.7, and 3.8 for pelvic tilt, rotation and obliquity, respectively. It is clearly seen that all of the possible solutions of rotation sequences result in nearly the same pelvis angles, in terms of both absolute values and trends.

Figure 3.6: Comparison of ZXY and Other Possible Sequences for Pelvic Tilt
Figure 3.7: Comparison of ZXY and Other Possible Sequences for Pelvic Rotation
Figure 3.8: Comparison of ZXY and Other Possible Sequences for Pelvic Obliquity
It is also clear from Figure 3.9 that flexion angles for hip joint are almost the same for all of the possible rotation sequences.

Figure 3.9: Comparison of ZXY and Other Possible Sequences for Right Hip Joint Flexion
Figure 3.10 illustrates the effects of all possible rotation sequences on hip internal rotation. Although similar trends are obtained for all possible rotation sequences, the absolute values are somewhat different for some rotation sequences.

Figure 3.10: Comparison of ZXY and Other Possible Sequences for Right Hip Joint Internal Rotation
Similar conclusions can be derived for the hip abduction, but to a lesser extent, as presented in Figure 3.11.

Figure 3.11: Comparison of ZXY and Other Possible Sequences for Right Hip Joint Abduction
It is also very clear from Figure 3.12 that knee joint flexion angles are almost the same for all of the possible rotation sequences, both in terms of values and trends.

Figure 3.12: Comparison of ZXY and Other Possible Sequences for Right Knee Joint Flexion
The internal rotation of knee joint, presented in Figure 3.13, are nearly the same up to the last portion of gait cycle, i.e. swing phase. In that region, the trends and values are different for some different sequences. Furthermore, for the swing phase, change in the direction of angles are observed in XZY and YZX sequences.

Figure 3.13: Comparison of ZXY and Other Possible Sequences for Right Knee Joint Internal Rotation
The knee valgus (abduction) appears to be the angle affected most by the rotation sequence as depicted in Figure 3.14. The values are similar at the mid region of gait cycle. But the region is too narrow. Most of the gait cycle is almost represented by two different curves, which implies that there are two kinds of solutions, while one of them abducts, the other adducts in the most of the gait cycle region.

Figure 3.14: Comparison of ZXY and Other Possible Sequences for Right Knee Joint Valgus

The percent differences associated with the angles are presented by Figures 3.15 to 3.23 in order to observe the extents at which different sequences would effect the KISS results. In order to compare the effect of rotation scheme on computed angles, the percent differences of the selected scheme with that of KISS (i.e. ZXY) are defined as in Equation 3.15

$$pd_i = 100 \times \frac{\theta_i - \theta_{i,KISS}}{ROM(\theta_{i,KISS})}$$

(3.15)

where $\theta_i$ is the angle of the scheme selected and $\theta_{i,KISS}$ is the angle calculated from
KISS. $ROM(\theta_{i,KISS})$ is the range of motion of the corresponding angle calculated by KISS. $i$ is the frame number. The ranges of angles are given in Table 3.2.

Table 3.2: Ranges of Angles

<table>
<thead>
<tr>
<th>Angle</th>
<th>Min</th>
<th>Max</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvic Tilt</td>
<td>1.2°</td>
<td>6.2°</td>
<td>7.4°</td>
</tr>
<tr>
<td>Pelvic Obliquity</td>
<td>−3.7°</td>
<td>4.7°</td>
<td>8.4°</td>
</tr>
<tr>
<td>Pelvic Rotation</td>
<td>−0.75°</td>
<td>7.7°</td>
<td>8.45°</td>
</tr>
<tr>
<td>Hip Flexion</td>
<td>−13.5°</td>
<td>27°</td>
<td>40.5°</td>
</tr>
<tr>
<td>Hip Abduction</td>
<td>−9°</td>
<td>0.86°</td>
<td>9.86°</td>
</tr>
<tr>
<td>Hip Rotation</td>
<td>1.7°</td>
<td>10.3°</td>
<td>8.6°</td>
</tr>
<tr>
<td>Knee Flexion</td>
<td>−4.6°</td>
<td>57°</td>
<td>61.6°</td>
</tr>
<tr>
<td>Knee Valgus</td>
<td>−15.8°</td>
<td>−2.9°</td>
<td>12.9°</td>
</tr>
<tr>
<td>Knee Rotation</td>
<td>−35.7°</td>
<td>−13.8°</td>
<td>21.9°</td>
</tr>
</tbody>
</table>

Percent differences of pelvic angles are presented in Figures 3.15, 3.16 and 3.17 for pelvic tilt, rotation and obliquity, respectively. The interpretation of pelvic angles being in good agreement with the KISS is justified by these difference plots. However, it is noticed that pelvic tilt has the most percent difference value of around 10% for the sequences of YXZ, YZX and ZYX. The other angles and other sequences of pelvic tilt have the percentages differences less than 1%.
Figure 3.15: Percent Differences between ZXY and Other Possible Sequences for Pelvic Tilt

Figure 3.16: Percent Differences between ZXY and Other Possible Sequences for Pelvic Rotation
The percentage differences are up to 4% for the hip flexion, as seen in Figure 3.18, while for most of the gait cycle differences are very low. The differences in hip flexion for different rotation sequences are observed to be maximum at the beginning and end of the swing phase.
Figure 3.18: Percent Differences between ZXY and Other Possible Sequences for Right Hip Joint Flexion

On the other hand, hip rotation and abduction, presented in Figures 3.19 and 3.20 respectively, have percentage differences up to 50% for three of the possible five alternative sequences. The other two sequences have nearly zero percentage differences. Also, it should be noted that in some portions there are very low percent differences.
As observed from Figure 3.21, except the sequence of ZYX, the knee flexion shows good agreement between the sequences with the differences not exceeding 5%. The ZYX sequence reaches 10% difference at the end of the gait cycle, while it is 5% or...
less for the most of the cycle.

![Graphs showing percent differences between ZXY and other possible sequences for Right Knee Joint Flexion.](image)

**Figure 3.21:** Percent Differences between ZXY and Other Possible Sequences for Right Knee Joint Flexion

Knee rotation angle, presented in Figure [3.22](#), have differences up to 60% for some of the sequences at certain instants of the gait cycle, approximately at the beginning and at the end of the swing phase, although for the most of the gait cycle, the differences are quite low. Percent variations in knee valgus for different rotation sequences are presented in Figure [3.23](#). Yet again, the knee valgus angle is the one most affected by the rotation sequences, in terms of percent differences too. The differences at certain instants of gait cycles are very high compared to all other angles for three of the five alternative sequences. The other two yield reasonably close results to that of the KISS for the whole gait cycle.
From all the figures presented above, it is clear that ZYX sequence yields very similar results to that of KISS for all of the angles considered. All of the percent differences between ZYX and ZXY (i.e. KISS results) are less than 10%. The possible reason for
that is the extraction of angles starts by flexion angles (in z-axis). The other similar sequence is XZY. The differences are very low except hip and knee internal rotations. One conclusion that may be derived from all these figures is that, for joints where only one of the possible three rotations is dominant, different sequences yield very close results. For example, in Table 3.2, ranges of hip flexion and knee flexion angles are considerably higher than the other two angles of these joints, resulting in insensitivity to the sequence used. This can also be seen mathematically from Equation (3.16), which defines a sequence in exponential matrix notation.

\[
\hat{C}^{(a,b)} = \hat{R}_i(\theta_1)\hat{R}_j(\theta_2)\hat{R}_k(\theta_3) = e^{\hat{\bar{u}}_{i}\theta_1}e^{\hat{\bar{u}}_{j}\theta_2}e^{\hat{\bar{u}}_{k}\theta_3}
\]  

(3.16)

Clearly, different sequences will result in similar angles if the two of the three rotations have much smaller magnitudes compared to the other rotation which is dominant. In the limiting case, the two angles being zero will result in a planar rotation, for which sequence is not an issue.
CHAPTER 4

ADDING SIMULATION FACILITY TO METU-KISS USING OPENSIM

OpenSim, as a free and open source software, allows the researchers to create libraries related to the laboratories, biomechanical models and experimental equipment to inverse kinematics and inverse dynamics calculations and simulate the gait. Furthermore, OpenSim also provides forward analysis, which enables to study the roles of muscles and assists surgery and treatment planning. In other words, mechanical outcomes of the surgery or treatment can be estimated on the OpenSim computer model by comparing the pre- and possible post-surgery/intervention. In addition to the individual or institutional advantage, the research collaboration is made possible by sharing the libraries developed on OpenSim.

In this chapter, the lower extremity model of KISS, which consists of pelvis, thighs, shanks and feet, will be simulated on the sub-model "Gait2392_Simbody.osim" supplied within the OpenSim 3.3 distribution. In Section 4.1, the basics of OpenSim simulation is presented together with the related terminology. Then, simulations of the KISS marker data will be performed and critically analyzed, using OpenSim graphical user interface (GUI) and METU-KISS GaitM Matlab codes with OpenSim library in Sections 4.2 and 4.3, respectively.

4.1 Overview of OpenSim

C++ is used for modeling and Java is used for GUI. The key functionalities of OpenSim are accessed through GUI. To perform a simulation, the first step is to scale the generic model to fit the subject dimensions. The scaling process may be manual by supplying the anthropometric data of the subject, or marker pairs may be used for
this purpose. Any two marker pair can be used to scale a segment in any of the three anatomical axes, transverse, medio-lateral and anterio-posterior. The scaling, either manual or by using marker pairs, is done by comparing the generic model dimensions with the subject specific data. In addition to scaling of dimensions, the mass properties including mass moment of inertia of bodies can also be scaled based on the same measurements. At the end, the OpenSim model will have the same geometric, mass and mass distribution properties as those of the subject.

Inverse kinematics is the second phase of the simulation. There are inevitable experimental errors that are violating the constraints of the joint models and rigidity assumption. OpenSim calculates model maker positions by solving a least squares problem in which any two markers on the same segment is minimized subject to constraints imposed by the joints and rigid body segments. In other words, the experimental marker locations and the model marker locations which are rigidly connected to the biomechanical model in the simulated gait are different.

The most valuable option of OpenSim is to use application programming interfaces (APIs) which allow the researchers use the available libraries to broaden the range of capabilities of OpenSim. API can be accessed through third-party programs such as C++, Matlab and Python. Development of APIs result in advancement in biomechanics research and sharing them. The necessary files to run a simulation, such as models and input files, is standardized, thus, it enables sharing models and data [55].

In Simbody upon which OpenSim is built, the term mobilizer is used to define the joint in the sense that motion specifications between the connected bodies and modelling requirements [56]. The mobilizers construct the body tree. In this tree, each body having a parent body is called child body. Except ground and the last body of the tree where the former cannot be a child and the latter cannot be a parent to any other body, each body has a parent and child. Different from the conventional notion which states joints remove some of the six DOFs, i.e. add constraints, mobilizers having certain limitations force the bodies to have the DOFs left from these limitations. A rigid body having a pin joint (or revolute joint) defined in the conventional notion has 6 ODEs for the total degrees of freedom of the body without the joint and 5 algebraic constraint equations due to the joint definition, adds up to 11 equations. In
the OpenSim definition of joints, one ODE is enough to model the system. Equations parametrizing the mobilizer are

1. homogenous transformation matrix, which is constructed by the rotation matrix and position vector of child frame $B$ with respect to parent frame $P$,

2. rotational and translational velocity of child frame $B$ with respect to parent frame $P$,

3. rotational and translational acceleration of child frame $B$ with respect to parent frame $P$,

4. the governing differential relationships relating the coordinate (such as angular position) to the mobility (to angular velocity).

The relationship between the bodies, both for the usage in GUI and API, is important. The bodies are referenced to each other as parent and child bodies. While the former is assumed to be fixed, the latter is moving relative to it. In Figure 4.1, the parent body $P$ (ground) is represented by the coordinate frames $P_0$ and $P$, while the child body $B$ is represented by coordinate frames $B_0$ and $B$. Assuming rigid bodies, the positions and orientations of $P$ with respect to $P_0$ and $B$ with respect to $B_0$ are fixed. The joint is defined by the position and orientation of $B$ with respect to $P$. Two coordinate frames are used for each body. The first one is fixed to body at certain position. The second one is also fixed to its body, however, the position and orientation with respect to the first frame is defined depending on the data collected. In this respect, the former frame represents anatomical reference while the latter stands for the technical one constructed by the markers.
The generic model of OpenSim is an extensible markup language (xml) in which rules are defined in terms of codes. The definitions include bodies, positions and orientations of them, mass properties, joints, allowed motions in the joints, markers, muscles and muscle forces. An xml file can be edited via an open-source code editor, such as Notepad++. The codes are readable by a human and a computer. The bodies are defined under the tag <BodySet>. Mass, mass center and inertia properties are defined under this tag. After the definition of a body, the joint relating it to a parent body, is defined by <Joint>. The joint types are weld, pin, slider, ball and free where there are zero, one rotational, one translational, three rotational and six DOFs, respectively. There is also another joint type called custom in which user may define the degrees of freedoms and the axes along or about which these DOFs are defined.

The original, sub-model of "Gait2392 _ Simbody" used in this study has the following bodies and properties defined by <BodySet> in the xml file. Although it is a musculoskeletal model, the details of muscles are beyond the scope of current study. The joint described below are the default joint types, and the user may introduce joint types other than these.

1. Ground: Only a parent body fixed on the laboratory frame.

2. Pelvis: The parent body is ground. The custom joint type has six DOFs with the
ground. Other than the regular rotational three DOFs of pelvis, there are three translational DOFs which simulate the translation of pelvis due to progression of gait, side swing and up-down motion.

3. Thighs: Both thighs have the similar properties. They (as childs) form joints with the pelvis (parent) having three rotational DOFs at hip joint centers.

4. Tibias: Both tibias have one rotational and two translational DOFs with respect to thighs (parents to tibias) in the right and left portion of the body. The rotational freedom in the knee joint is flexion/extension angle. The two translational freedoms are defined as functions of flexion/extension angle along x (anteroposterior) and y (superoinferior) axes. Thus, tibias include one independent degree of freedom.

5. Feet: Talus, calcaneus and toes are treated as different bodies. There are two rotational degrees of freedom. Toes and calcaneus can be considered fixed to each other. Calcaneus can rotate with respect to talus and talus can rotate with respect to tibia. Thus, child-parent relationships exist between calcaneus-talus and talus-tibia.

The markers are defined under the <MarkerSet> tag. A marker is defined by a body on which it is attached and the location in the local frame of the body. The Helen-Hayes marker set applied on the generic model is shown in Figure 4.2. The marker positions in this generic model is obtained by cadaveric studies.

---

1 Note that there exist another marker (15) which is not visible in Figure 4.2.
Table 4.1 shows markers, their placements and the bodies on which markers are attached. The coordinates x, y and z are the coordinates of markers in the local frames of the bodies given for further references. LHEE, LTOE, RHEE and RTOE markers are actually placed on the body called calcaneus. Calcaneus is the one of the three bodies constructing the foot complex.
Table 4.1: Helen-Hayes Marker Set Applied to the Model

<table>
<thead>
<tr>
<th>No</th>
<th>Marker</th>
<th>Body</th>
<th>x</th>
<th>y</th>
<th>z</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>LASI</td>
<td>Pelvis</td>
<td>12.23</td>
<td>11.85</td>
<td>-123.10</td>
</tr>
<tr>
<td>1</td>
<td>RASI</td>
<td>Pelvis</td>
<td>12.23</td>
<td>11.85</td>
<td>123.10</td>
</tr>
<tr>
<td>4</td>
<td>LTHI</td>
<td>Left Thigh</td>
<td>19.33</td>
<td>-254.98</td>
<td>-70.66</td>
</tr>
<tr>
<td>6</td>
<td>LKNE</td>
<td>Left Thigh</td>
<td>10</td>
<td>-396.31</td>
<td>-58.33</td>
</tr>
<tr>
<td>8</td>
<td>LTIB</td>
<td>Left Tibia</td>
<td>14.44</td>
<td>-278.94</td>
<td>-46.75</td>
</tr>
<tr>
<td>12</td>
<td>LANK</td>
<td>Left Tibia</td>
<td>-7</td>
<td>-409.19</td>
<td>-48.94</td>
</tr>
<tr>
<td>10</td>
<td>LHEE</td>
<td>Left Foot</td>
<td>-16.37</td>
<td>17.50</td>
<td>0</td>
</tr>
<tr>
<td>14</td>
<td>LTOE</td>
<td>Left Foot</td>
<td>180</td>
<td>12.5</td>
<td>0</td>
</tr>
<tr>
<td>3</td>
<td>RTHI</td>
<td>Right Thigh</td>
<td>19.33</td>
<td>-254.98</td>
<td>70.66</td>
</tr>
<tr>
<td>5</td>
<td>RKNE</td>
<td>Right Thigh</td>
<td>10</td>
<td>-396.31</td>
<td>58.33</td>
</tr>
<tr>
<td>7</td>
<td>RTIB</td>
<td>Right Tibia</td>
<td>14.44</td>
<td>-278.94</td>
<td>46.75</td>
</tr>
<tr>
<td>11</td>
<td>RANK</td>
<td>Right Tibia</td>
<td>-7</td>
<td>-409.19</td>
<td>48.94</td>
</tr>
<tr>
<td>9</td>
<td>RHEE</td>
<td>Right Foot</td>
<td>-16.38</td>
<td>17.5</td>
<td>0</td>
</tr>
<tr>
<td>13</td>
<td>RTOE</td>
<td>Right Foot</td>
<td>180</td>
<td>12.5</td>
<td>0</td>
</tr>
<tr>
<td>15</td>
<td>SACR</td>
<td>Pelvis</td>
<td>-153.49</td>
<td>28.85</td>
<td>0</td>
</tr>
</tbody>
</table>

Anatomical local frames of bodies are aligned with the global (laboratory) frame axes in the neutral position. The origin of pelvic local frame is at the midpoint between the anterior superior iliac spines. The femoral heads are chosen as the origins of local frames of thighs on both sides. The right and left tibias have the origins of their local frames at right and left knee joint centers, respectively. The knee joint center is defined at the neutral position where knee flexion is zero. While talus local frame origins at the ankle joint, that of calcaneus is at the heel. The heel position considered here is different than the heel marker, rather it is near to the plantar surface of the foot.

The local anatomical frames are oriented such that x-, y- and z- axes are along the anterior and superior directions and to the right of the subject, respectively. y-axis is the long axis of the thigh and tibia. The solution method for the angles is body-fixed XYZ angle sequence, called rotated frame based XYZ. Mathematically, it is equivalent to initial frame based ZYX angle sequence. The angles and translations (coordinates in general) describing the lower extremity model are described in Table 4.2.
### Table 4.2: Coordinate Information of the Model

<table>
<thead>
<tr>
<th>Coordinate</th>
<th>Body/Joint</th>
<th>Axis (x-y-z)</th>
<th>Range (rad or m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tilt</td>
<td>Pelvis</td>
<td>0 0 1</td>
<td>-0.35 0.35</td>
</tr>
<tr>
<td>List (Obliquity)</td>
<td>Pelvis</td>
<td>1 0 0</td>
<td>0 0.35</td>
</tr>
<tr>
<td>Rotation</td>
<td>Pelvis</td>
<td>0 1 0</td>
<td>-0.26 6.28</td>
</tr>
<tr>
<td>Translation x</td>
<td>Pelvis</td>
<td>1 0 0</td>
<td>No Constraint</td>
</tr>
<tr>
<td>Translation y</td>
<td>Pelvis</td>
<td>0 1 0</td>
<td>No Constraint</td>
</tr>
<tr>
<td>Translation z</td>
<td>Pelvis</td>
<td>0 0 1</td>
<td>No Constraint</td>
</tr>
<tr>
<td>Flexion</td>
<td>Hip</td>
<td>0 0 1</td>
<td>-2.09 2.09</td>
</tr>
<tr>
<td>Abduction</td>
<td>Hip</td>
<td>1 0 0</td>
<td>-2.09 2.09</td>
</tr>
<tr>
<td>Rotation</td>
<td>Hip</td>
<td>0 1 0</td>
<td>-2.09 2.09</td>
</tr>
<tr>
<td>Knee angle</td>
<td>Knee</td>
<td>0 0 1</td>
<td>-2.09 0.17</td>
</tr>
<tr>
<td>Translation 1</td>
<td>Knee</td>
<td>1 0 0</td>
<td>f(Knee Angle)</td>
</tr>
<tr>
<td>Translation 2</td>
<td>Knee</td>
<td>0 1 0</td>
<td>f(Knee Angle)</td>
</tr>
<tr>
<td>Ankle Angle</td>
<td>Ankle</td>
<td>-0.11 -0.17 0.98</td>
<td>-1.57 1.57</td>
</tr>
<tr>
<td>Subtalar Angle</td>
<td>Subtalar</td>
<td>0.79 0.61 -0.12</td>
<td>-1.57 1.57</td>
</tr>
<tr>
<td>Mtp Angle</td>
<td>Mtp Joint</td>
<td>-0.58 0 0.81</td>
<td>-1.57 1.57</td>
</tr>
</tbody>
</table>

* As a function of knee angle
** Metatarsophalangeal

The red, yellow and green colored axes in Figure 4.3 are x-, y- and z-axes of pelvis, respectively. The walking direction is along x-axis in the neutral position. The pelvic angles of tilt, obliquity (called list in OpenSim) and rotation are around z-, x- and y-axes, respectively.

![Figure 4.3: Pelvic Frame](image)

There are three translations, x, y and z, of pelvic frame with respect to the laboratory fixed frame. These translations are responsible for the articulation of the lower extremity, and have no anatomically meaningful correspondences. Hip joint angles,
flexion, abduction and rotation are defined as the rotations of thigh local frame with respect to pelvic frame. The axes in Figure 4.4 are in the same alignment with the pelvic frame. The hip joint angles are obtained as rotations around this frame where flexion, abduction and rotation are around z (green), x (red) and y (yellow) axes.

![Image](image.png)

**Figure 4.4: Thigh Local Frame and Hip Joint Angles**

Since knee is modeled as a hinge, the only knee angle is the knee flexion occurring around z-axis. The difference between the normal hinge joint in mechanics is that there are two translations along x- and y-axes of tibia with respect to those of thigh, specified as functions of knee flexion angle. Therefore, these translations are not considered as degrees of freedom. The splines describing the change of translations with respect to the angle is defined in the XML file, which depends on the work by Delp [57]. Figure 4.5 shows the child (tibia) and parent (thigh) frames of the knee joint.
The ankle joint complex includes four bodies, tibia, talus, calcaneus and toes. It is called ankle joint complex since it is formed by three different joints. Three joints forming the ankle joint complex are called ankle, subtalar and metatarsophalangeal joints between talus and tibia, calcaneus and talus, calcaneus and toes, respectively. These angles are measured as rotations about unit vectors defined in Table 4.2 which are also shown in Figure 4.6. Based on Delp [57], ankle angle is defined around an axis, which is close to z-axis. The subtalar angle in OpenSim, the rotation of calcaneus relative to talus, is around an axis nearly in x-y plane. Metatarsophalangeal (mtp) angle, defined between the toes and the calcaneus, is around an axis obtained by rotating z axis by 8° around the vertical axis.
4.2 Simulation of Gait using OpenSim GUI

Once the bodies and markers are defined in the xml file of the model, there are three more steps for simulation, namely marker trajectories, scaling and inverse kinematics. First, (.trc) file that contains the marker trajectories must be constructed. This is done by a script written in Matlab R2018a. The marker names are the same as those in model xml file. A model is loaded as shown in Figure 4.7.

![Figure 4.7: Loading a model in GUI](image)

The second step is called scaling. In this step, the anthropometric parameters, like mass, inertia and segment length, of the generic model are scaled to obtain the subject-specific parameters. It is possible to use different scaling parameters in different axes.
Scaling is done by averaging the first 20 frames of data. Although there exists an option for scaling with static shot data, the trials have revealed that it may result in some unrealistic stature and segment dimensions. The marker sets used in the scaling procedure are summarized in Table 4.3. The dimensions and inertia parameters of a body is scaled based on the corresponding measurements of “Marker1” and “Marker2” on the generic model and experiment. According to Table 4.3, the bodies are scaled in all principal (global) axes with the same scale parameter, but an option to use different scales along different directions is available.

Table 4.3: Marker Pairs Used in Scaling the Generic Model

<table>
<thead>
<tr>
<th>Body</th>
<th>Axes</th>
<th>Marker1</th>
<th>Marker2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td>x-y-z</td>
<td>RASI</td>
<td>LASI</td>
</tr>
<tr>
<td>Right Femur</td>
<td>x-y-z</td>
<td>RASI</td>
<td>RKNEE</td>
</tr>
<tr>
<td>Left Femur</td>
<td>x-y-z</td>
<td>RASI</td>
<td>LKNEE</td>
</tr>
<tr>
<td>Right Tibia</td>
<td>x-y-z</td>
<td>RKNEE</td>
<td>RANK</td>
</tr>
<tr>
<td>Left Tibia</td>
<td>x-y-z</td>
<td>LKNEE</td>
<td>LANK</td>
</tr>
<tr>
<td>Right Talus</td>
<td>x-y-z</td>
<td>RKNEE</td>
<td>RANK</td>
</tr>
<tr>
<td>Left Talus</td>
<td>x-y-z</td>
<td>LKNEE</td>
<td>LANK</td>
</tr>
<tr>
<td>Right Calcaneus</td>
<td>x-y-z</td>
<td>RHEEL</td>
<td>RTOE</td>
</tr>
<tr>
<td>Left Calcaneus</td>
<td>x-y-z</td>
<td>LHEEL</td>
<td>LTOE</td>
</tr>
<tr>
<td>Right Toes</td>
<td>x-y-z</td>
<td>RHEEL</td>
<td>RTOE</td>
</tr>
<tr>
<td>Left Toes</td>
<td>x-y-z</td>
<td>LHEEL</td>
<td>LTOE</td>
</tr>
</tbody>
</table>

The next step towards simulation of gait in OpenSim environment is to determine the weightings of markers, to be used in least squares minimization to nearly satisfy rigidity condition subject to joint constraints, as described in Section 2.4 and expressed mathematically in Equation 2.25. As a first attempt, unit weightings are assigned for all markers, assuming that all markers are equally reliable. This marker set weightings will be referred to as M1. The second weighting scheme, M2, is based on findings of a previous work [58], and are given in Table 4.4.
In what follows, simulations for M1 and M2 marker weightings sets are performed in OpenSim by using experimental marker locations obtained from a previously conducted test of a normal male subject with existing KISS hardware and software, and the results are presented in Figures 4.8, 4.9 and 4.10 for three representative angles. Although results are obtained for all kinematic variables during gait, these three angles presented are deemed sufficient to draw conclusions relevant to marker weightings. In these simulations, the default joint model of the OpenSim (labelled as LL) which considers the knee as a single degree-of-freedom joint with dependent translations are used. The effect of different joint models will be subsequently analyzed.
KISS results are also included in the comparisons as they have been previously compared with the established studies on gait analysis and very good agreement have been reported [59]. Comparison between OpenSim and KISS results needs to be done based on the general trend, peaks and valleys, their timing, and peak-to-peak values, rather than the absolute values. This is because, OpenSim model markers and experimental markers are different, and this causes a constant shift in angles, as it happens even with the same equipment, protocol and software with differences in palpating places where markers are to be placed.

Figure 4.8: Pelvic Obliquity Angles Obtained from KISS and OpenSim for M1 and M2 marker weightings sets

Pelvic obliquity obtained from M1 and M2 scaling and IK scheme are shown in Figure 4.8 together with KISS results. Both M1 and M2 yield peak-to-peak values close to the normal range. M2 appears to have similar characteristics with KISS, while M1 gives somewhat a different trend.
The knee flexion angle is shown in Figure 4.9. The peak-to-peak values of both M1 and M2 are consistent with that of the KISS, and within the normal range. However, M1 characteristics appear to be closer to KISS results than M2 characteristics.

The largest differences between the results are observed for hip internal rotation angle, presented in Figure 4.10. The peak-to-peak values of both M1 and M2, but especially that of M1, are out of normal range. Since there exists no generally accepted pattern for the relatively little variation of hip internal rotation angle, there is no point to compare the results based on the shapes of the curves.

Based on these observations, it is concluded that neither the unity weightings (M1) nor the weightings of another study (M2) yield satisfactory results with OpenSim, and therefore, it is necessary to develop new set of weightings.
For this purpose over 20 different sets of marker weightings were tested, and a third marker weightings set (M3) is generated, which also includes additional markers other than the ones used in the experimental protocol. These are the right and left joint centers for hip (RHJC and LHJC), knee (RKJC and LKJC) and ankle (RAJC and LAJC) joints. Another difference of M3 from M1 and M2 is that wand markers (LTHI, LTIB, RTHI and RTIB) placed on the right and left tibia are excluded from the scaling weightings, as suggested by Kainz et al. [58]. The weightings of M3 thus obtained are also presented in Table 4.4.

The marker weighting set M3 is compared with M1 and M2 for all 11 joint angles of the lower extremity. Compared with the results of M1 and M2, M3 results are observed to be closer to those of the KISS, and peak-to-peak values remain within normal bands, except hip rotation angle for which there exists no generally accepted pattern. Therefore, it was decided to continue the analysis with M3 marker weighting.
In addition to marker weighting sets, the joint models also affect the kinematic results. Effect of joint models on kinematics of gait is investigated by considering the following joint models:

1. LL: The default joint models, i.e. three rotational freedoms for hip and ankle, and single-rotational freedom (flexion) with two dependent translations for the knee joints.
2. 3-DOF: This is the joint models used in KISS, i.e. all three joints have three rotational degrees of freedom.
3. 3-6-3: Three rotational degrees of freedoms for hip and ankle joints, and six degrees of freedoms for the knee joint which allows, in addition to three rotations, three tibiofemoral (between tibia and femur) translations.

Compared with the default model (LL), 3-DOF model, improved both variations and peak-to-peak values of joint angles. This is illustrated for hip rotation in Figure 4.11 and for hip flexion in Figure 4.12 in a sense they are nearly in the normal bands. 3-DOF model results in hip joint angle variation corresponding to normal bands, except between 70-80% of gait cycle (Figure 4.11). In Figure 4.12, difference in hip flexion for LL and 3-DOF models appear to be slight, but a very characteristic peak and valley within the first and last 10% of the gait cycle is obtained by 3-D model. There is an improvement gained by 3-DOF model.
Figure 4.11: Hip Rotation Angles Obtained from KISS and OpenSim
In Figures 4.13, 4.14, 4.15, 4.16, and 4.17, 3-DOF model is compared with 3-6-3 model for four different angles. Variation of pelvic obliquity within gait cycle are pretty close to each other for the two models compared (Fig. 4.13). There appears to be a vertical shift between them, and both give less peak-to-peak value than the KISS result. These two are sufficient to continue with 3-DOF model. The last model 3-6-3, on the other hand, is superior to 3-DOF model. Firstly, it is clearly seen that the pelvic obliquity of 3-DOF model is out of normal bands. 3-6-3 model have pelvic obliquity angles are not also similar in variation but also comparable in variation. These are presented in Figure 4.13.
Figure 4.13: Pelvic Obliquity Angles Obtained from KISS and OpenSim

Hip abduction pattern of 3-6-3 model is clearly closer to KISS results and yields peak-to-peak value within normal band, when compared to the variation with 3-DOF model, which, especially at the beginning and the end of the gait cycle, yield some divergence from normal pattern (Figure 4.14).
Similar observations are applicable for hip internal rotation (Fig. 4.15) and knee internal rotation (Fig. 4.16). 3-6-3 model have the hip rotation values in the normal bands. Also, the magnitudes and variations are less in 3-6-3 model, which represents characteristics of this angle better. Yet again, 3-DOF model yields some divergence from the normal pattern towards the end of the gait cycle. In Figure 4.16 peak-to-peak value of knee internal rotation of KISS reach some unrealistically high values, and the same are observed for both 3-DOF and 3-6-3 models. It is believed that the existing systems without cameras located at the top of the laboratory yield such abnormalities in both hip and knee internal rotations.
Figure 4.15: Hip Rotation Angles Obtained from KISS and OpenSim
Knee valgus (or abduction) angle (Fig. 4.17) have values somewhat out of normal ranges for KISS and 3-DOF models. The smooth and less variation of 3-6-3 model is more reasonable than others.
These tests indicate that M3 marker weighting set and 3-6-3 joint model yield more reasonable angles compared with other combinations of sets and models. Moreover, it is known that some relative translations at the knee (between femur and tibia) exist during normal gait, and only 3-6-3 model yield such knee translations. For the gait test considered in this study, 3-6-3 model and M3 weighting scheme gave 25 mm anterior-posterior, 4 mm medio-lateral, and 16 mm superior-inferior knee translations. Li et al. [24] showed that differences on joint translations exist between the measuring systems, and radiography is the only reliable method. They reported knee translations up to 15 mm during gait. In this regard, the 3-6-3 model and M3 weighting scheme may also be considered acceptable in terms of knee joint translations.

It should, however, be noted that these limited tests are far from being conclusive, and more research is needed in this regard.

In what follows foot angles obtained from OpenSim with 3-6-3 joint model and M3
marker weighting set are compared with those of the KISS.

The foot is defined as three bodies as stated in Section 4.1. The angles related to the foot were stated as ankle, subtalar and mtp angles. First of all, it is seen that the axis defined in Section 3.3, which is related to the foot dorsiflexion angle, may be obtained by the ANK axis given in Figure 4.6. It was stated that dorsiflexion occurs around an axis which is near to z axis of global frame. ANK axis seems to meet this requirement.

![Figure 4.18: Ankle (OpenSim) and Dorsiflexion (KISS) Angles](image)

The great agreement of ankle angle of OpenSim and dorsiflexion of KISS can be seen in Figure 4.18. That means the ANK axis defined in OpenSim greatly replaces the dorsiflexion axis.

Another angle calculated by OpenSim is subtalar angle. In the corresponding Sections, it was stated that subtalar angle takes place around an axis which is in the x-y plane with very small contribution of z-axis of global frame. Since the foot rotation
angle is the rotation of foot around the shank y-axis which is mainly in global y-axis but may have x components during walking, it might be expected that subtalar angle can represent the foot rotation of the KISS protocol. Figure 4.19 shows that this expectation holds. It should be noted that rotation angle, labeled as KISS, is the negative of angle calculated by OpenSim. The negative of an angle means nothing but the axis of rotation is reversed.

![Subtalar and Foot Rotation Angles](image)

Figure 4.19: Subtalar and Foot Rotation Angles

Kinematics of foot are defined in various different ways in different gait analysis systems and protocols. Despite the differences in OpenSim and KISS protocols, the two-angle definitions of foot kinematics coincided unexpectedly well.

4.3 Simulation of Gait using OpenSim by Matlab Interface

OpenSim modifies the experimental marker coordinates in order to nearly satisfy the rigid body assumption of segments in the least squares sense, and also to satisfy the
constraints imposed by the joints, and comes up with a new set of marker coordinates
called model markers. The purpose of this section is to analyze the extent with which
OpenSim modifies various experimental markers. However, this analysis cannot be
readily performed by OpenSim GUI described in the previous sections. Therefore,
Kiss-GaitM, Matlab codes that process the marker and force plates data of KISS sys-
tem as described in Chapter [3] is extended to run a simulation through OpenSim
API. In this way, the simulation and results can be visualized and total squared error
and marker RMS errors can be analyzed for different scaling parameters and mod-
els. Here, "marker error" is defined as the difference between the experimentally
measured marker locations and the marker locations obtained by the process of error
minimization of OpenSim.

RMS calculation procedure is given in Equation [4.1] where n is the number of markers
included in the analysis, the caret (^) indicates the model marker location in the
corresponding axis.

\[
RMSE_x = \sqrt{\frac{\sum_{i=1}^{n} (\hat{x}_i - x_i)^2}{n}}
\]
\[
RMSE_y = \sqrt{\frac{\sum_{i=1}^{n} (\hat{y}_i - y_i)^2}{n}}
\]
\[
RMSE_z = \sqrt{\frac{\sum_{i=1}^{n} (\hat{z}_i - z_i)^2}{n}}
\]
\[
RMSE = \sqrt{RMSE_x^2 + RMSE_y^2 + RMSE_z^2}
\]

In Table [4.3] the effect marker weightings are presented in terms of maximum, RMS
and total squared marker error values, for the most affected marker indicated in the
last column.

Table 4.5: Marker Errors with Default Model (LL) For Different Marker Weightings
Set

<table>
<thead>
<tr>
<th>Scaling</th>
<th>Total Squared Error</th>
<th>RMS</th>
<th>Max</th>
<th>Name</th>
</tr>
</thead>
<tbody>
<tr>
<td>M1</td>
<td>63.89</td>
<td>65.26</td>
<td>142.43</td>
<td>LTIB</td>
</tr>
<tr>
<td>M2</td>
<td>7.25</td>
<td>25.68</td>
<td>59.62</td>
<td>SACR</td>
</tr>
<tr>
<td>M3</td>
<td>11.18</td>
<td>25.64</td>
<td>71.71</td>
<td>SACR</td>
</tr>
</tbody>
</table>
All error measures are greatly improved with weighting sets M2 and M3. This is due to the fact that wand markers used in M1 are not related to bone anatomy, and are prone to soft tissue movements more than the markers placed on palpable anatomical landmarks. On the other hand, all marker error values are pretty close to each other for M2 and M3 sets. This indicates that differences in joint angles due to M2 or M3 sets arise in the inverse kinematics phase, rather than the scaling phase.

Table 4.6: Marker Errors for Different Models with the Same Scaling Method M3

<table>
<thead>
<tr>
<th>Model</th>
<th>Total Squared Error</th>
<th>RMS</th>
<th>Maximum</th>
</tr>
</thead>
<tbody>
<tr>
<td>LL</td>
<td>11.18</td>
<td>25.64</td>
<td>71.71</td>
</tr>
<tr>
<td>3-DOF</td>
<td>10.88</td>
<td>25.29</td>
<td>72.00</td>
</tr>
<tr>
<td>3-6-3</td>
<td>9.97</td>
<td>24.22</td>
<td>74.05</td>
</tr>
</tbody>
</table>

Effects of different joint models are presented in Table 4.6. Here, M3 scaling method is used where joint centers are utilized and wand markers (RTHI, RTIB, LTHI and LTIB) are excluded from the scaling weightings list but added to inverse kinematics calculations. Therefore error values given in Table 4.6 for different joint models are all for the most affected marker, SACRUM. The total squared errors, RMS and maximum errors are comparable in all of the joint models. In the preceding section gait kinematics were shown to be different for different joint models. It is therefore concluded that joint model employed has very little effect on scaling (model marker locations), but has considerable effect on inverse kinematics.

In table 4.6 slight difference is noted for 3-6-3 model. This is due to the fact that 6-DOF knee joint model relieves all the constraints for this joint, and OpenSim minimization algorithm needs to deal with constraints at the hip and ankle joints only. At this point, the question of “what would happen to marker errors (difference between experimental and model markers) and resulting inverse kinematic results (joint angles) if all constraints in all joints are removed?

To investigate the answers to the above question, a new 6-DOF model is generated on the OpenSim in which hip and knee have 6, and ankle has 3 degrees-of-freedoms.

As expected, joint angles obtained with 6-DOF model are closer to that of the KISS,
since the model markers are closer to experimental markers due to removal of some of the constraints at the hip joint which do not exist in KISS. However, rigid body conditions and ankle joint constraints still exist, and some marker errors are expected to occur.

Figures 4.20, 4.21, 4.22 and 4.23 compares marker errors for three models, LL, 3-DOF, and 6-DOF; along certain laboratory axes for five representative markers, namely, LASI, RASI, SACRUM, LKNE and RKNE. Note that laboratory x-axis is along the direction of progression, y-axis is upwards, and z-axis perpendicular to both, and all models use M3 scaling weights.

Similar errors are obtained for x- and z- component of markers RASI, LASI and SACR. Small differences are observed between the models where 6-DOF model produces the least amount of error most of the time. The maximum errors are around 5 mm (Figure 4.20)

![Figure 4.20: RASI, LASI and SACR x-axis Errors for Different Models](image)

Marker errors appear to be affected more by the joint model along y-direction (Figure
The default model LL and 3-DOF model have similar characteristics, but model 3-DOF yield smaller errors. 6-DOF model, on the other hand, have the errors nearly zero throughout the time interval considered which includes one complete gait cycle. These results are consistent with the expected outcome of released joint constraints.

![Figure 4.21: RASI, LASI and SACR y-axis Errors for Different Models](image)

Horizontal axes are times in seconds  
Vertical axes are marker errors in mm

Figure 4.21: RASI, LASI and SACR y-axis Errors for Different Models

Similar observations can be done for knee markers in Figures 4.22 and 4.23. Differences in the scales of the figures must be taken into account when knee marker errors are compared with pelvis markers. Specifically, the knee marker errors in y-axis are higher than x- and z-axes, unlike the pelvic markers. This is attributed to experimental errors, which are unavoidable due to insufficiency of experimental setup, accuracy and resolution of cameras, calibration and linearization processes and the process of 3-D coordinate determination from 2-D camera images.
Figure 4.22: RKNE and LKNE y-axis Errors for Different Models

Horizontal axes are times in seconds
Vertical axes are marker errors in mm

Figure 4.23: RKNE and LKNE z-axis Errors for Different Models

Horizontal axes are times in seconds
Vertical axes are marker errors in mm
Another observation on Figures 4.20-4.23 is that, for a specific marker maximum errors occur at similar timings for all models and directions. This is clarified in Figure 4.24 which shows maximum error (in mm) of the right knee joint center with time (in seconds). Clearly maximum error occurs at the heel strike. Therefore impulsive actions during gait such as the heel-strike cause considerably larger experimental errors than relatively smoother actions during gait. This explains why, despite the removal of joint constraints, marker errors still exist.

![Figure 4.24: Right Knee Joint Center Marker Errors for 6DOF Model](image)

In conclusion, marker errors are due to experimental errors violating rigidity condition and joint constraints. When joint constraints are relieved, marker errors become less and remaining errors can be attributed to experimental errors.

Relieving joint constraints affect not only marker errors and inverse kinematic results (joint angle variations) but also give rise to relative translations at the joint. Knee joint translations of 3-D model were previously discussed in Section 4.2. The 6-D model
introduce translational freedoms at the hip joint as well. Akiyama et al. [60] studied hip motion by magnetic resonance imaging and found out that the hip translations are present up to 1 mm. As hip joint translations (translations of femoral head with respect to acetabulum cavity) are very limited, validity of the 6-D model needs to be checked in terms of resulting hip joint translations.

Figure 4.25 shows resulting hip joint translation along z-axis of the laboratory frame. There exists almost 9 mm relative translation along this direction (medio-lateral). Hip joint translations along anterior-posterior and superior-inferior directions reach 5 mm. Clearly these are unrealistic relative translations between femoral head and acetabulum cavity.

![Figure 4.25: Right Hip Joint Center Translation in z-axis](image)

The translational motion between the femur and tibia at the knee joint along superior-interior (y-axis) direction is given in Figure 4.26. The amount of translation reaches up to 12 mm.
Figure 4.26: Right Knee Joint Center Translation in y-axis

Knee joint translation along anterior-posterior (x-axis) direction is presented in Figure 4.27 which shows a maximum of almost 25 mm relative translation. Both of these translations along y- and x-axes are somewhat higher than expected from a normal subject.
The knee joint translation along medio-lateral (z-axis) direction has the smallest value (3.5 mm), and is within the normal range. It should be noted that relative translations at the knee joint are very close to other for 3-6-3 and 6-DOF models.

It may be concluded that using 6-DOF model is not anatomically (and physically) meaningful. Considering both the joint angles and translations in terms of ranges of motion, normal bands and trend of lines, 3-6-3 model and M3 weightings scheme appear to be the most appropriate strategy among the alternatives considered in this study.
Inertial measurement units (IMUs) are called 9-axis sensors since a unit is composed of gyroscope, accelerometer and magnetometer each sensing corresponding components in 3 principal local axes of the sensor unit. Advances in the sensor technology may lead to the usage of IMUs in the gait analysis due to less equipment requirement, high frequency data acquisition and out of laboratory usage. This chapter is devoted to the use of IMUs in kinematic data acquisition and reconstruction of 3D human gait on computer using OpenSim. OpenSim, however, relies on markers. Therefore, it is required to use application programming interface (API) of OpenSim, with access to its libraries.

In Section 5.1 in order to gain insight on the basics of OpenSim API, free response of a double pendulum to initial conditions is simulated, and results are compared with numerical solution of equations of motion of the double pendulum. Secondly, OpenSim is tested, this time with artificial IMU data obtained from the transformation matrices of the double pendulum. Next in Section 5.2 features of IMU data is explained together with its problems. A detailed IMU processing algorithm is presented in Section 5.3. Finally, tests are performed with a simple pendulum and gait trials in Section 5.4.

5.1 Simulation of Double Pendulum

There are two reasons of choosing a double pendulum as a starting point. First of all, creating a model and performing a simulation on OpenSim is experienced and
documented for further studies. The second reason is that, double pendulum is one of the simplest examples consisting of rotation of bodies with respect to ground and each other.

Double pendulum consists of two massless rods. Two point masses are attached to the end of both rods and its motion under gravity is examined. First, the governing equations of motion are derived and their numerical solution is presented. Next, OpenSim API is utilized to construct the bodies and simulate the motion, and the results are compared.

5.1.1 Double Pendulum Dynamics

Double pendulum consisting of two rods and two spheres as point masses is shown in Figure 5.1. The angles are as defined in OpenSim. The lengths of the massless rods are \( h_1 \) and \( h_2 \), and masses of the point masses are \( m_1 \) and \( m_2 \), respectively. The equations of motion are derived using Lagrange’s Equation.

![Double Pendulum](image)

Figure 5.1: Double Pendulum

The general form is given in Equation 5.1 where the equation is written for each body \( i \) and the Lagrangian \( L \) is the difference between kinetic and potential energies. The
The x- and y-coordinates of spheres can be obtained from Equation 5.2, and given in complex numbers as in Equation 5.2.

\[
Z_1 = h_1 e^{i(\theta_1 + \pi/2)}
\]

\[
Z_2 = h_1 e^{i(\theta_1 + \pi/2)} + h_2 e^{i(\theta_1 + \pi/2 + \theta_2)}
\]  

The x- and y-coordinates of spheres can be written from Equation 5.2 and given in Equation 5.3.

\[
x_1 = -h_1 \sin \theta_1 \\
y_1 = h_1 \cos \theta_1
\]

\[
x_2 = -h_1 \sin \theta_1 - h_2 \sin(\theta_1 + \theta_2) \\
y_2 = h_1 \cos \theta_1 + h_2 \cos(\theta_1 + \theta_2)
\]

The kinetic and potential energies of the system are presented in Equation 5.4.

\[
T = \frac{1}{2} m_1 (\dot{x}_1^2 + \dot{y}_1^2) + \frac{1}{2} m_2 (\dot{x}_2^2 + \dot{y}_2^2)
\]

\[
V = m_1 g y_1 + m_2 g y_2
\]

After the necessary manipulations, the equations for the first and second point masses are presented in Equations 5.5 and 5.6.

\[
\begin{bmatrix}
    h_2^2 m_2 + h_1^2 (m_1 + m_2) + 2h_1 h_2 m_2 \cos \theta_2 \\
    h_2 m_2 (h_2 + h_1 \cos \theta_2)
\end{bmatrix}
\begin{bmatrix}
    \dot{\theta}_1 \\
    \dot{\theta}_2
\end{bmatrix}
= g [h_1 (m_1 + m_2) \sin \theta_1 + h_2 m_2 \sin(\theta_1 + \theta_2)] + (\dot{\theta}_1 + \dot{\theta}_2) h_1 h_2 \dot{\theta}_2 \sin \theta_2
\]

\[
(h_2 m_2 (h_2 + h_1 \cos \theta_2)) \dot{\theta}_1 + h_2^2 \ddot{\theta}_2 = -h_2 m_2 (-g \sin(\theta_1 + \theta_2) + h_1 \dot{\theta}_1^2 \sin \theta_2)
\]  

The Equations 5.5 and 5.6 are arranged in the matrix form as follows,

\[
\begin{bmatrix}
    h_2^2 m_2 + h_1^2 (m_1 + m_2) + 2h_1 h_2 m_2 \cos \theta_2 & h_2 m_2 (h_2 + h_1 \cos \theta_2) \\
    h_2 m_2 (h_2 + h_1 \cos \theta_2) & h_2^2
\end{bmatrix}
\begin{bmatrix}
    \dot{\theta}_1 \\
    \dot{\theta}_2
\end{bmatrix}
= \begin{bmatrix}
    g [h_1 (m_1 + m_2) \sin \theta_1 + h_2 m_2 \sin(\theta_1 + \theta_2)] + (\dot{\theta}_1 + \dot{\theta}_2) h_1 h_2 \dot{\theta}_2 \sin \theta_2 \\
    -h_2 m_2 (-g \sin(\theta_1 + \theta_2) + h_1 \dot{\theta}_1^2 \sin \theta_2)
\end{bmatrix}
\]

With the initial positions of \( \theta_1 = \pi \) and \( \theta_2 = \pi/2 \) and zero initial velocities, the equations are numerically solved on Matlab, and the results are presented in Figure 5.8.
for 10 seconds of simulation. The parameters are taken as $m_1 = 30$ kg, $m_2 = 50$ kg, $h_1 = 0.5$ m and $h_2 = 0.7$ m.

![Figure 5.2: Double Pendulum Results by Matlab](image)

5.1.2 Double Pendulum Simulation on OpenSim-API

OpenSim structure consists of layers. The ground layer is a computational layer denoted by "SimTK" on which OpenSim is built. The computational capabilities, listed on the right hand side of Figure 5.3 such as optimizer and integrator, can all be used within OpenSim API. The second layer is the modelling as denoted by "Model". In the modelling layer, model components are defined to form a complete model by its components. The equation of motion of a multi-body system is based on the model components. All the modelling choices such as DOFs of bodies and constraints are expressed in this layer. Next layer is "Analysis" composed of "Modeler", "Solver" and "Reporter". In "Modeler" part, an existing model is generally modified depending on the problem. In the context of gait analysis, this modification is the scaling of the model to obtain subject specific model. In the second part, "Solver", equations of motions of the system (or the model) are solved to obtain kinematics or kinetics. The examples are the Inverse Kinematics and Inverse Dynamics to determine body and joint angles and joint moments, respectively. "Reporter" class is just responsible
to report or display the results. The "Application" layer is above the API and consists of GUI and plug-ins, which may include new classes to extend the capabilities of OpenSim, created by API. The architecture is presented in Figure 5.3.

![OpenSim API Architecture](image)

**Figure 5.3: OpenSim API Architecture**

Double pendulum example is constructed by C++ programming medium. Model is constructed using “Model” class of OpenSim. The visualizer can be set to be true to view the simulation of motion without using OpenSim directly. After defining the gravity with a 3-D vector for the model, ground body is specified. A display can be added to visualize the ground. Each rod and point mass in the pendulum is considered as one body. “Body” class is utilized to construct the rods with the masses of spheres having the mass center at the sphere center. Therefore, the rods are realized as massless. The display geometries are added for the bodies in the “Body” class. The body needs its mass, mass center and mass moment of inertia to be specified. Since the double pendulum example consists of point masses, the mass moment of inertia about mass centers are taken as zero. After defining the bodies, the two pin joints are to be defined. Since there exists predefined “PinJoint” class in OpenSim, definition of joints need only joint locations and orientations in child and parent bodies. The defined joint types are free, ball, ellipsoid, gimbal, planar and slider. Additionally, another joint definition included is called "custom joint" allowing the user to define different joint types with different allowed degrees of freedoms. For example, a joint having one degree of freedom and two other motions related to this DOF can only be
defined by "custom joint". Finally, the range of motions for the joints are specified, and the model is updated with the bodies. Since the free response of the double pendulum under the gravity is analyzed, Simbody is utilized in this step. Runge-Kutta-Merson integrator implementation is triggered by defining the initial pose of the system. The integration up to 10 seconds is performed with the accuracy of $10^{-4}$ degrees. The differences between the two results obtained in the previous section and here are very low when compared to the range of motions. The maximum differences for $\theta_1$ and $\theta_2$ are 0.21 and 0.36 degrees, respectively. This is a negligible difference caused by different numerical methods used. The accumulation of differences are presented in Figure 5.4.

![Figure 5.4: The Differences between OpenSim and Matlab Results](image)

### 5.1.3 Artificially Created IMU Driven OpenSim-API Simulation

The main goal of this chapter is to utilize OpenSim for IMU based gait analysis. IMUs are used to obtain orientation information of bodies on which they are attached. It is assumed that two IMUs are attached on both masses in planar motion. The transformation matrices of the spheres with respect to the ground can readily be obtained from the solution of equations of motion of Section 5.1. These matrices are treated as if they are obtained from the attached ideal IMUs, and subsequent steps of simulation...
are performed. The transformation matrices for two spheres are as follows,

\[
M_1 = \begin{bmatrix}
\cos \theta_1 & \sin \theta_1 & 0 \\
-\sin \theta_1 & \cos \theta_1 & 0 \\
0 & 0 & 1
\end{bmatrix}
\]

(5.8)

\[
M_2 = \begin{bmatrix}
\cos(\theta_1 + \theta_2) & \sin(\theta_1 + \theta_2) & 0 \\
-\sin(\theta_1 + \theta_2) & \cos(\theta_1 + \theta_2) & 0 \\
0 & 0 & 1
\end{bmatrix}
\]

(5.9)

The transformation matrices in Equations 5.8 and 5.9 are for the rotations around z-axis only.

In the OpenSim part, C++ codes are to be implemented using OpenSim libraries. Before implementation, input transformation matrices are defined in the same file format (.trc) of markers. Since it is not possible to input transformation matrices to OpenSim directly, each row of matrices is treated as marker coordinates. Thus, a marker file consisting of 6 markers is constructed. The C++ coding part includes the placement of virtual orientation sensors to the segments. It is similar to the virtual markers used in the generic model for the marker based analysis. The virtual sensor placement is performed at the upright position of double pendulum. The axes of sensors are aligned with the axes shown in Figure 5.1. An orientation class is defined in the C++ code which takes the model, sensor output (as transformation matrices) and the body names as inputs. The constructor (a special function) of the class assigns the inputs to the objects (variables) of the class. In the "OrientationSensors" class, sensors and observations are defined. As the names imply sensors stand for the IMUs attached and observations are the sensor outputs. The bodies are then assigned to the sensors. Any orientation between the sensor and its corresponding body can be specified. It is worthy to note that the observations are with respect to the ground fixed laboratory frame. This was implied in Equation 5.9. The transformation matrices of each sensor is obtained from the input trc file. For every frame, each transformation matrix is assigned to corresponding observation to change the orientation of its body.

Figure 5.5 compares time variations in two angles during free response obtained from artificially created IMU driven OpenSim codes, and from the Matlab solution of equations of motion. The same result is expected since the joints are purely revolute and
no error is present. The differences are in the order of 0.4 degree at maximum, which is caused by round-off errors.

Figure 5.5: Double Pendulum Angles from Matlab and Opensim by IMUs

5.2 IMU Measurements and Related Problem

The inertial measurement unit (IMU) consists of gyroscope, accelerometer and magnetometer to sense the angular velocity and linear acceleration of sensor frame and earth magnetic field, respectively. In recent years, using IMUs in the gait analysis has attracted great deal of attention due to its advantages such as low cost, compact size and portability which leads to outdoor usage. Main disadvantage, on the other hand, is sources of errors.

In this study a low cost IMU, LSM9DS1 by SparkFun, is used. The histogram and Gaussian distribution of biased noise of this device is presented in Figure 5.6 which is taken for 5 seconds from a stationary sensor unit.
The noise will result in accumulation of error in the sensor orientation estimation. This phenomenon is known as integration drift. For a sensor standing still, gyroscope readings in all axes must be zero. Since the angular rate is integrated to obtain angular position, the integration of zero will result in zero. That is, the orientation of sensor remains the same. If there are small readings in the sensors, the integration of them leads to a different orientation. At each time instant, a small contribution is added to the previous one, this accumulates into measurable magnitudes in time.

A test is conducted when the aforementioned IMU sensor is stationary. Gyroscope readings (angular velocities) along x-, y-, and z-axes are recorded, and integrated independently of each other for 5 seconds (corresponding to 500 frames with 100 Hz of frequency) to obtain angular displacement drifts about these axes. Figure 5.7 shows the resulting drifts. The drift in y-axis may reach up to 14°. Angular drifts about x- and z-axes have the maximum drifts of 3° and 2°, respectively in 5 seconds.
To overcome the drift problem, first the bias is removed by subtracting the mean value of the stationary value from all the data readings, i.e. by offsetting the data. Secondly, Butterworth filter is applied to the raw signal to get rid of the noise and smooth the data. Butterworth filter is widely used in the human movement data acquisition [61, 62]. Butterworth filter of different orders and cut-off frequencies are tested, and a sixth-order Butterworth filter with a cut-off frequency of 6 Hz, with filtering in forward and backwards manner to prevent any phase lag or lead, is found to give the best results among the alternatives tried.

After applying the offsetting and filtering described above, the integration drift presented in Figure 5.7 is reduced to that shown in Figure 5.8.
5.3 Mathematical Description of IMU Processing

IMUs employed in gait analysis can be divided into two groups. In the first group, the manufacturers, such as Xsens, supply the measurement units along with a processing software to calculate the orientation of the sensor or joint angles, directly [61]. However, only extension-flexion angles are provided. The other group use different algorithms to calculate sensor’s orientation while implementing their own kinematic models [62] or using synthetic data [63]. In this work, an IMU processing software is developed and applied for an off-the-shelf equipment based on the mathematical description given by [64].

While obtaining angular drifts of Figure 5.7, the angular velocity readings are directly integrated by assuming that angular motion takes place around each axis separately. In other words, motion around each of the x-, y- and z-axes occur separately. This obviously is not the case for 3-D motion which includes simultaneous rotations around each axis. To express the mathematical relationship between the orientation angles and the angular rates, it is essential to define the reference frames as a sensor frame ($\hat{\mathbf{F}}_s$) and inertial frame ($\hat{\mathbf{F}}_i$), indicated by s and i as subscripts, respectively. As stated previously, the angular velocity of sensor frame with respect to the inertial frame is
sensed and expressed in sensor frame, given in Equation \[5.10\]

\[
\hat{\omega}_{s/i}^{(s)} = \begin{bmatrix}
\omega_1 \\
\omega_2 \\
\omega_3
\end{bmatrix} = \hat{C}_{s,i}^{(s,i)} \hat{\omega}_{s/i}^{(i)}
\]  

(5.10)

where \(\omega_1, \omega_2\) and \(\omega_3\) are the measured velocity components. The angular velocity expressed in inertial frame can be expressed using the exponential representation of the transformation matrix in Equation \[5.11\] in 3-2-1 (or Z-Y-X) sequence.

\[
\hat{C}_{(i,s)}^T = \hat{C}_{(s,i)} = e^{-\tilde{u}_1\phi} e^{-\tilde{u}_2\theta} e^{-\tilde{u}_3\psi}
\]  

(5.11)

The yaw, pitch and roll angles are defined by \(\psi, \theta\) and \(\phi\), respectively. The differentiation with respect to time in the inertial frame results in Equation \[5.12\].

\[
\hat{\omega}_{s/i}^{(i)} = \dot{\psi}\tilde{u}_3 + \dot{\theta}\tilde{u}_2 + \dot{\phi}\tilde{u}_1
\]  

(5.12)

Combining equations \[5.10\], \[5.11\] and \[5.12\] to obtain the measured quantities in terms of orientation angles, Equation \[5.13\] is obtained.

\[
\hat{\omega}_{s/i}^{(s)} = \begin{bmatrix}
\omega_1 \\
\omega_2 \\
\omega_3
\end{bmatrix} = \begin{bmatrix}
1 & 0 & -\sin \theta \\
0 & \cos \phi & \cos \theta \sin \phi \\
0 & -\sin \phi & \cos \theta \cos \phi
\end{bmatrix} \begin{bmatrix}
\dot{\phi} \\
\dot{\theta} \\
\dot{\psi}
\end{bmatrix}
\]  

(5.13)

Since it is sought to obtain the rates of orientation angles for integration, the Equation set \[5.14\] is more useful.

\[
\dot{\phi} = \omega_1 + \omega_2 \sin \phi \tan \theta + \omega_3 \cos \phi \tan \theta \\
\dot{\theta} = \omega_2 \cos \phi - \omega_3 \sin \phi \\
\dot{\psi} = \omega_2 \tan \theta + \omega_3 \cos \phi \sec \theta
\]  

(5.14)

However, integration of Equation set \[5.14\] does not produce satisfactory results due to drift. Therefore, an algorithm which fuses the information gained from accelerometer and magnetometer with the gyroscope is needed for a better estimate of the angular positions. The sensor fusion technique adopted in this work has weightings associated with gyroscope, accelerometer and magnetometer readings, and weighs the gyroscope reading most in the dynamic motion region. In other words, the orientation is obtained mainly from accelerometer and gyroscope readings when there is nearly no motion, such as during the static shot.
The algorithm is presented using quaternion representation. The quaternion forms of the sensed quantities are expressed in Equation Set 5.15 where subscripts $\omega$, $a$ and $m$ are for gyroscope, accelerometer and magnetometer, respectively. The quaternions are named as $s$ to imply that the quantity is expressed in the sensor frame.

\[
\begin{align*}
    s_\omega &= \begin{bmatrix} 0 & \omega_x & \omega_y & \omega_z \end{bmatrix} \\
    s_a &= \begin{bmatrix} 0 & a_x & a_y & a_z \end{bmatrix} \\
    s_m &= \begin{bmatrix} 0 & m_x & m_y & m_z \end{bmatrix}
\end{align*}
\]  

(5.15)

It is required to have the reference quaternions for acceleration and magnetic field vector expressed in the earth fixed reference frame, which are denoted by $e$ in Equation set 5.16.

\[
\begin{align*}
    e_a &= \begin{bmatrix} 0 & 0 & 0 & 1 \end{bmatrix} \\
    e_m &= \begin{bmatrix} 0 & \sqrt{m_x^2 + m_y^2} & m_z & 0 \end{bmatrix} = \begin{bmatrix} 0 & m_2 & 0 & m_4 \end{bmatrix}
\end{align*}
\]  

(5.16)

It is assumed that the acceleration exerted on the bodies is just the gravitational acceleration. Using primarily the gyroscope readings in the high acceleration regions justifies this assumption. Besides, there is the magnetic field vector that helps reducing the effects of dynamic conditions characterized by high accelerations. The quaternion representation of the orientation of Earth frame with respect to the sensor frame is denoted by $q_{E,S}$. The rate of this quaternion is expressed as in 5.17.

\[
\dot{q}_{E,S} = \frac{1}{2} q_{E,S} \times s_\omega
\]  

(5.17)

Using the gradient descent algorithm, the quaternion is estimated in two steps. First of all, in Equation 5.19, the quaternion rate at the current step, denoted by $\dot{q}_{E,S}^t$, is calculated from the quaternion rate calculated from the gyroscope data, denoted by $\dot{q}_{E,S}^{\omega,t}$, and the information coming from the accelerometer and magnetometer readings processed with the gradient descent algorithm. The gradient is calculated as in Equation 5.18. The $|\nabla f|$ term in Equation 5.19 is the norm of this gradient.

\[
\nabla f = J^T f
\]  

(5.18)

\[
\dot{q}_{E,S}^t = \dot{q}_{E,S}^{\omega,t} - \beta \frac{\nabla f}{|\nabla f|}
\]  

(5.19)

In the second step, the numerical integration takes place. For the discrete data, the integration is carried out by trapezoidal rule presented in Equation 5.20.

\[
q_{E,S}^t = q_{E,S}^{t-1} + \dot{q}_{E,S}^t \Delta t
\]  

(5.20)
Returning back to Equation 5.19 an objective function vector, \( f_a \), can be defined as the difference between the sensed acceleration and transformation of earth’s gravitational acceleration to the sensor frame as given in Equation 5.21 with the quaternion product. The similar objective function vector, \( f_m \), is obtained for the magnetometer. The magnetometer reading and earth’s magnetic field direction expressed in the sensor frame are subtracted from each other to get a function that should be zero. The latter objective function vector related to the magnetometer is given in Equation 5.21.

\[
\begin{align*}
\mathbf{f}_a &= q_{E,S}^a \mathbf{e}_a \mathbf{q}_{E,S}^s - s_a \\
\mathbf{f}_m &= q_{E,S}^m \mathbf{e}_m \mathbf{q}_{E,S}^s - s_m
\end{align*}
\] (5.21)

Cascading these two vectors, function \( f \) in Equation 5.19 is obtained. Objective function error, \( \nabla f \), is calculated from the function itself and its Jakobian, \( J \). After this definition, the meaning of \( \beta \) in Equation 5.19 can easily be given. It may be called the error in the gyroscope measurement, or gyroscope based orientation information, that is to be minimized by the direction based on the objective function error term.

\[
J = \begin{bmatrix}
\frac{\partial f_1}{\partial q_1} & \frac{\partial f_2}{\partial q_1} & \cdots & \frac{\partial f_n}{\partial q_1} \\
\vdots & \vdots & & \vdots \\
\frac{\partial f_1}{\partial q_n} & \frac{\partial f_2}{\partial q_n} & \cdots & \frac{\partial f_n}{\partial q_n}
\end{bmatrix}
\] (5.22)

With the definition of Jacobian given in Equation 5.22 where \( n = 4 \), each objective function’s Jacobian are taken separately. The naming convention of a quaternion is given by \( q_{E,S} = [q_1 \ q_2 \ q_3 \ q_4] \) since its elements are to be used in the objective functions and Jacobian. Applying the necessary quaternion algebra, the objective functions given in 5.21 takes the matrix form of Equations 5.23 and 5.24 for the accelerometer and magnetometer, respectively. While computing the objective functions, the definitions given in Equations 5.15 and 5.16 are utilized. The last step is to take the Jacobian of each objective function with the definition given in Equation
Referring back to Equation $5.19$, the derivative of the quaternion that describes the sensor frame’s orientation with respect to the earth frame orientation can be calculated. The last step is to take numerical integral of each element to calculate the quaternion.

The obtained quaternions from the algorithm are converted into transformation matrices to feed them into OpenSim. As it is done in the double pendulum example, the transformation matrices defining the bodies with respect to the fixed reference frame enables OpenSim to extract the angles depending on the constraints existing on the model.

### 5.4 Experimental Results

The subroutines written for double pendulum of Section 5.1 aimed to test whether OpenSim is able to process IMU data or not. With the IMU processing algorithm described in Section 5.3, real the experiments can now be carried out and analyzed on OpenSim-API. In this section, results are presented for a simple pendulum test and for a gait experiment using off-the-shelf IMUs.
Three tests are carried out on a simple 1-DOF pendulum for three different orientations of the IMU unit attached on the pendulum. In each test, pendulum is released from rest at approximately $45^\circ$ from its static equilibrium position under the action of gravity. In the first test, the IMU is aligned with the long axis of the pendulum, which results in parallel IMU-fixed and pendulum-fixed frames. In the second test, the IMU is rotated around the rotation axis of the pendulum by $45^\circ$. In this way, the only common axis between pendulum and IMU fixed frames is the rotation axis. In third test, the IMU is rotated around one of its axes other the rotation axis of the pendulum, to obtain an arbitrary orientation of IMU with respect to the pendulum.

Results of the first and the second tests displays the expected damped oscillations with a certain frequency, as illustrated in Figure 5.9, for the first test.

![Figure 5.9: The Rotation of a Simple Pendulum](image)

However, the algorithm developed did not work for the third test. This situation can be explained as follows: In the first and second tests, since the axis of rotation of the pendulum coincides with one of the axes of the IMU unit, there is only one dominant angular velocity reading to be integrated, while in the third test IMU velocity readings exist along each direction. This is obviously a major shortcoming of the algorithm developed, and a remedy would be to employ more sophisticated but costly filtering methods, such as Kalman filter.

On the other hand, high frequencies and high accelerations involved the pendulum test
is far from being representative of frequencies and accelerations involved in human gait. Therefore, the developed algorithm is tested with a gait experiment utilizing IMU’s for motion capture.

IMUs are attached to the lower limb segments. In OpenSim part, the C++ codes are written to define the segments and OpenSim model sensors (in analogy with the model markers). The transformation matrices obtained from the IMU processing algorithm developed is supplied to OpenSim API, which completes the simulation process as described earlier.

The flexion angle of the knee joint is presented in Figure 5.10 which has similar characteristics obtained from marker based analysis for a typical gait trial as explained in Chapters 3 and 4. It should, however, be noted that the camera based and IMU based gait experiments could not be carried out on the same subject, because at the time of conducting IMU based test, we experienced a technical problem in the KISS software. Therefore, results presented in Chapters 3 and 4 are from an earlier test conducted on a different subject.
The pelvic angles and hip joint angles are presented in Figures 5.11 and 5.12, respectively. Their variation and values are also in a good agreement with a typical gait pattern.
Figure 5.11: Pelvis Angles Obtained From IMU and OpenSim
Figure 5.12: Hip Joint Angles Obtained From IMU and OpenSim
CHAPTER 6

CONCLUSION AND FUTURE WORK

6.1 Conclusion

This thesis deals with the improvements of the gait analysis system of METU, called KISS. The study can be divided into three main parts. The first and second parts are related to the kinematics of a typical gait analysis conducted by the camera-based KISS gait analysis system. First, the effects of different rotation sequences in joint angle calculations are analyzed, and secondly, an open-source software, OpenSim is implemented to introduce simulation capability to the existing system. The third part of the thesis is devoted to a different measurement technique using inertial measurement units (IMUs), as an alternative to the existing camera-based measurements.

After reviewing the literature on the kinematics of the gait analysis, effects of different joint angle calculation methods by using different Euler/Cardan angle sequences were investigated. Looking from the robotics perspective, determination of joint angles using different rotation sequences will yield different results. However, for a typical normal gait, it is shown that certain joint angles remain the same even if the rotation sequence is altered. Pelvic, hip flexion and knee flexion angles are shown to be nearly the same for different rotation sequences, where percentage differences do not exceed 10%.

The second part of the work is the simulation of gait using OpenSim. The marker set of the model is adjusted based on the marker set employed in the KISS system. The default lower extremity model in the OpenSim distribution 3.3. is chosen as the starting point. In addition to this model, two other joint models are generated. The
first one includes 3 degrees of freedom at each joint of the lower extremity without allowing any joint translations. In the second model, the knee joint is free to translate in every direction which corresponds to 6 degrees of freedom at the joint. Besides, different marker weighting schemes were tested in comparison with normal gait pattern, and an appropriate combination of a joint model and a marker weighting set were chosen for further study.

Several codes were developed for using smoothed and drift-free IMU data in OpenSim simulation environment. The first code was written for a double pendulum, which uses the orientation matrices of the pendulum with respect to an earth fixed reference frame as input. Subsequently, the code was further developed to accept artificially created IMU data in the form of orientation matrices as input. Then, an algorithm for processing the IMU data and obtaining the orientation information in terms of quaternion was presented. The performance of the algorithm was tested using a simple pendulum experiment. Three test conditions were considered. In the first test, IMU sensor axes and pendulum axes were aligned. In the second test, the IMU was rotated with respect to the pendulum around the axis of rotation of the pendulum. The last test was conducted by rotating the second IMU orientation with respect to one of the other two axes of the pendulum. In this way, an arbitrary orientation of IMU with respect to the pendulum was obtained. The IMU processing algorithm worked well for the first two tests but failed for the last test. This is attributed to the relatively high frequency of pendulum compared to a normal human gait. Therefore, it was decided to continue with the algorithm which is computationally less expensive compared to Kalman filtering algorithms used in conventional IMU processing.

Finally, human gait data acquired by IMUs were processed by the algorithm, and transformation matrices of the segments involved in the analysis were obtained. Then, the simulation was performed by codes written to enable OpenSim use transformation matrices instead of marker positions.

The right and left portions of the body were treated as two branches. Then, each branch may be thought as pendulums in series. By inputting the orientation matrices of each segment in each branch, the code performed the simulation. The kinematic results agreed well with the literature on the normal human gait.
6.2 Future Work

In this thesis, different rotation sequences of joints are tested for a typical gait data. It will be more informative if effects of rotation sequences are investigated for a much wider range of gait data representing age and gender difference, and also certain pathological conditions.

Additional work is needed on the implementation and usage of OpenSim for the usage in METU KISS-Gait system. First, the inverse dynamics, i.e. obtaining the joint reaction forces and moments, must be implemented. To begin with, the assumption made by METU KISS-Gait software packages could be maintained, that the foot is rigid and the ground reaction forces and moments acting on it are directly transmitted to ankle joint. Next, the assumption may be questioned by applying the ground reactions directly on the foot. Another development on OpenSim implementation in KISS Gait system would be to include calculation and reduction of what is known as “residual”, which is defined as the difference between the measured ground reaction forces and the multiplication of total mass and mass center acceleration of the whole body model. In other words, during gait Newton’s Second Law for the whole body is not satisfied due to measurement and data processing errors, and residual is a quantitative description of all these errors. In literature, this inconsistency is solved by adjusting the center of gravity of the whole body and its acceleration. Finally, incorporation of muscles into the gait analysis would be very useful, especially for the diagnosis and treatment or surgery assessment of pathological gait. There are several ways which muscles are incorporated in gait analysis. For example, in Computed Muscle Control (CMC) each modeled muscle excitation is predicted by using optimization techniques involving the kinematics and kinetics calculated beforehand.

In conclusion, after kinematics implemented in this thesis, future work may involve inverse dynamics, residual reduction and CMC on Matlab and/or OpenSim.

The last recommendation of future work is related to the IMU implementations. First of all, additional work is needed on the drift of the sensors. Although drift is lowered to a certain level by the applied filters, it may be helpful to implement algorithms with more effective filtering capabilities. One example is the Kalman filter which is computationally more expensive than the algorithm used in this work, but it is more
powerful in the long term due to its updates on the parameters depending on the estimates. Different versions of Kalman filters may be applied to improve the performance. Additionally, after processing of IMU data, the OpenSim part of the work may be enhanced by introducing an optimization process for the input orientation information, similar to the minimization algorithm used in OpenSim for experimental markers. In other words, the output of IMUs may be processed by an error minimization algorithm. Lastly, a proper scaling method is needed to determine anthropometric parameters of the subject. In OpenSim the generic model is scaled for the subject by using marker coordinates, which are missing in IMU setup. One alternative would be to employ ultrasonic markers during the stand still position of the subject, and use these static shot marker data in OpenSim for scaling of the generic model for a specific subject.
REFERENCES


