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Development and Validation of a Three-Dimensional Biomechanical Model of the Lower Extremity

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1. Introduction

One of the important applications of the computer modelling of human body is the area of joint replacement where a validated model can be used for surgery planning. It is known that the evolution of total knee and total hip replacement has been influenced to a great extent by the knowledge obtained from gait analysis studies (Andriacchi and Hurwitz, 1997). Many of the mechanical problems associated with these devices have been evaluated in terms of the mechanics of walking where the magnitude and pattern of the forces at the hip and knee joints obtained from gait analysis studies have been used as design criteria of both total hip and total knee replacements.

Gait analysis provides a unique opportunity to obtain objective information that cannot be obtained through other clinical means (Andriacchi and Hurwitz, 1997). For instance, several investigators have advocated the use of gait analysis for planning surgery and therapy treatments for children with cerebral palsy (Lofterod, et al., 2007; Kay, et al., 2000; Molenaers, et al., 2006). Improvement in gait after multi-level surgery using kinematic data has been documented, while kinematics provides information on dynamic joint motion kinetics is essential for differentiating between primary deformities and secondary responses.

The potential benefits of gait analysis are improved treatment decision making, so that surgery and other treatments result in improved walking capability. Also, the information generated from the gait analysis of patients with total joint replacements has been utilized as a tool for assessing recovery following these procedures, where the key to the analysis of functionality following joint replacement is the ability to identify the adaptations corresponding to the joint design features.

It is very difficult to determine muscle force/power output from multiple muscles simultaneously without affecting the pattern of normal movements (Naganoa, et al., 2005). Fortunately, computer modeling can provide useful insights for human biomechanics. Most in-vivo experiments only reveal the forces in the joint and not the surrounding muscle forces or their point of application. It is also known that finding the internal forces in the body by in-vivo experiments alone is difficult and sometime impossible. Because of the inherited redundancy in the musculoskeletal system (Crowninshield and Brand, 1981b) a desired motion can be achieved by an infinite

number of activation patterns of the muscles. Because of the additional forces induced by muscles, knowledge of how the muscle forces are applied can be very important for understanding the bone strength and degeneration around the replacement so that replacement joints can be designed and tested with the loading environment in which they will operate.

A number of studies have been conducted to investigate the forces in the different joints in the lower extremities. For instance, with regard to hip joint, a number of studies have been conducted to validate musculoskeletal models using both instrumented hip replacements (Bergmann et al., 2001, Brand et al., 1994) and the activity levels of muscles obtained by means of electromyography (Crowninshield et al., 1978). Others have studied the forces produced in the knee joint (Piazza et al., 1996), and the ankle joint as well (Orendurff et al., 2002).

While these studies have shown that the hip contact forces and muscle activity levels can be represented by computational methods, several theories have been developed (Crowninshield et al., 1978, Rasmussen et al., 2001), but the question of how the body recruits the muscles for a given activity needs more understanding. This question can be better answered through the use of 3D musculoskeletal modeling. Using three-dimensional computer modelling makes it possible to evaluate the behaviour of individual muscles during various human movements (Eltoukhy & Asfour, 2009 and Yamaguchi, 2001).

Although the mechanics of muscles and joints easily become statically indeterminate, which means that there are not enough equilibrium equations describing the muscle activation available to resolve the forces in the system. Another complication is caused by the muscles in the system because they can only pull and some muscles contribute to the joint movements. This constrains the space of possible solutions and adds a fair bit of complexity to the problem.

In principle, resolving forces is a question of setting up the equilibrium equations and solving them. But in biomechanics in particular, there are several complications. There have been several optimization methods used to solve this, the basic requirement to the Inverse Dynamic Analysis solver is that it must be able to cope with both statically indeterminate problems and unilateral forces elements.

1.1 Knee replacement

According to the American Academy of Orthopaedic Surgeons, there are about 270,000 knee replacement operations performed each year in the United States. Although about 70% of these operations are performed in people over the age of 65, a growing number of knee replacements are being done in younger patients. Knee replacement is basically a surgery for people with severe knee damage during which the surgeon removes damaged cartilage and bone from the surface of the knee joint and replaces them with a metal and plastic surface (Figure 1).

In other words, it involves the resurfacing of the worn out parts of the knee using a metal component on the end of the femur and the top of the tibia, with a plastic bearing in between. The total knee replacement had been studied in the literature by a large number of researchers. One of the most relevant studies to our research work was the study conducted by Benedetti et al (2000). In their study the authors determined the gait function as it relates to the residual quadriceps' strength and to the specific components of the quadriceps removed in patients treated with total knee replacement.

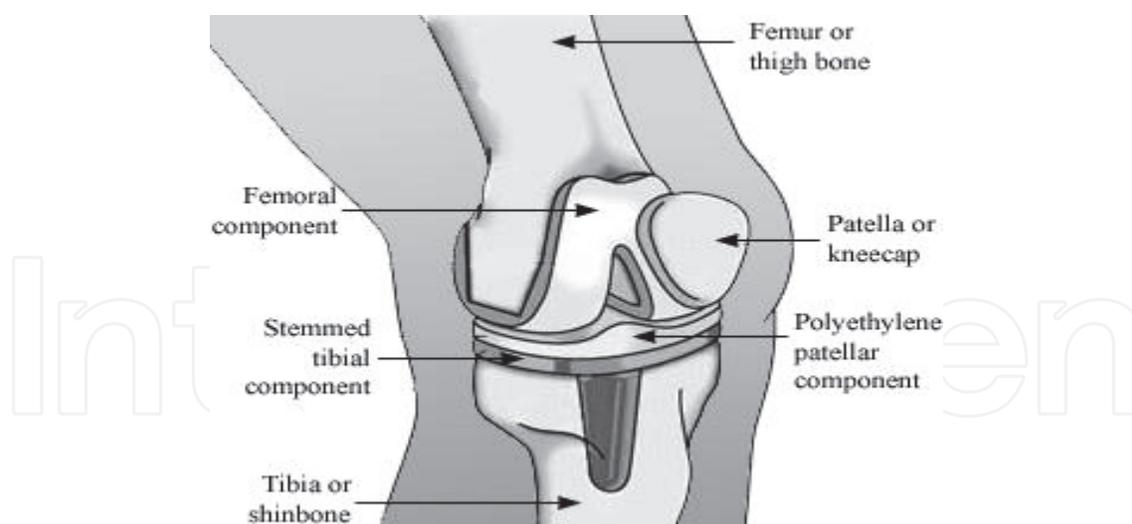


Fig. 1. Total knee replacement surgery (Healthbase)

2. Musculoskeletal modeling

Three-dimensional (3D) computer modeling possesses the advantage of providing useful insights and allowing the 3D-evaluation of the behavior of individual muscles during the different human movements. As it has been shown by a number of researchers such as Alkjaer et al. (2001), 3D vectors better represent the line of action of the different muscles as compared to two-dimensional (2D) vectors specially when investigating the location of both the origin and insertion of muscles.

One of the important applications of the computer modeling of human body is the area of joint replacement where a validated model can be used for instant surgery planning. It is known that the evolution of total knee and total hip replacement has been influenced to a great extent by the knowledge obtained from gait analysis studies.

Many of the mechanical problems associated with these devices have been evaluated in terms of the mechanics of walking where the magnitude and pattern of the forces at the hip and knee joints obtained from gait analysis studies have been used as design criteria of both total hip and total knee replacements. Because of the additional forces induced by muscles, knowledge of how the muscle forces are applied can be very important for understanding the bone strength and degeneration around the replacement so that replacement joints can be designed and tested with the loading environment in which they will operate.

2.1 Model development

One of the main features of the approach used in this study is the possibility to drive a 3D musculoskeletal model entirely from the motion capturing data and to be able to predict the muscle recruitment pattern required to perform a certain task, in this chapter the task studied was a full gait cycle. The gait data collected is used basically for model validation and that is by comparing the data obtained versus the database Hip98 developed by Bergmann et al. (1998).

2.1.1 Subjects and procedures

Five healthy subjects aged 25 ± 2 years participated in this study. The participants were given instructions including an explanation of the test procedures, proper attire, and the

expected duration of the testing. The data was collected at the University of Miami Biomechanics Research Laboratory, USA, with an approval from the Institutional Review Board (IRB) at the University of Miami. Forty eight reflective markers were placed at the different land marks (e.g. joints center lines and segments). First, a static trial is conducted where the patient stands at a T- pose; the goal of this static trial is to use it for labelling purposes. Then five dynamics trials were performed. The subject was instructed to walk at his/her normal speed across the laboratory and on top of four force plates (Figure 2).

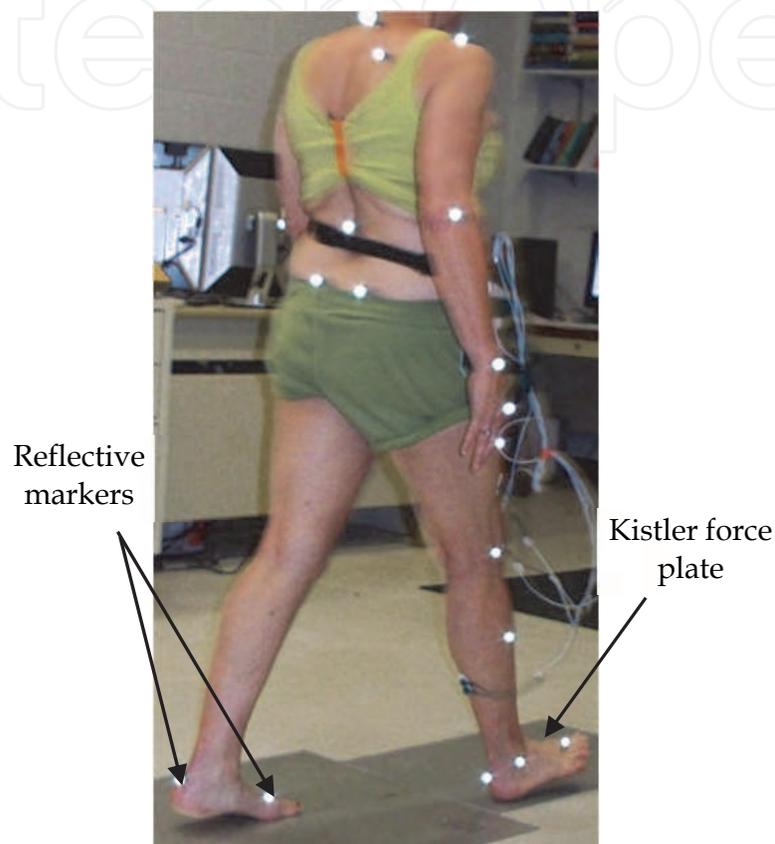


Fig. 2. Actual subject walking across the lab during a gait motion capturing session.

2.1.2 Instrumentation and data collection

The laboratory incorporates a ViconNexus® Motion Capturing System (Oxford Metrics, United Kingdom). The motion capturing system integrates and synchronizes four Kistler force plates (Model: 9253B, sampling rate: 2400 Hz), and eight MX cameras. The MX cameras provide 1024 x 1024 pixel resolution and frame rates up to 250 Hz. The setup including the force plates and the MX cameras is shown in figure 3.

The MX cameras capture the reflected infrared light from the markers placed on the subject's skin and thus the X, Y, and Z coordinates of the body segments and joints at the different time steps are recorded continuously. The reconstructed data output of the motion capturing session is shown in figure 4, which depicts the stick figure of the subject's lower extremity, the reflective markers as recorded by the cameras, the segments' center lines and reference frames, and the ground reaction vectors acting on the subject.

As shown in figure 4, the data collected from the motion capturing (Mocap) system is first reconstructed and each marker is labeled in order to identify the different body segments

and joints. Then the segments' center lines are determined accordingly at the same time the synchronized force plates' data is recorded. Body segmental parameter values were derived from Horsman et al. (2007). Hip joints were modelled as universal joints where hip joint flexion/extension, adduction/abduction and internal/external rotation were allowed. Knee joints were modelled as hinge joints, while ankle joints were modelled as biaxial joints, ankle joint dorsi/plantar flexion and inversion/eversion were allowed.

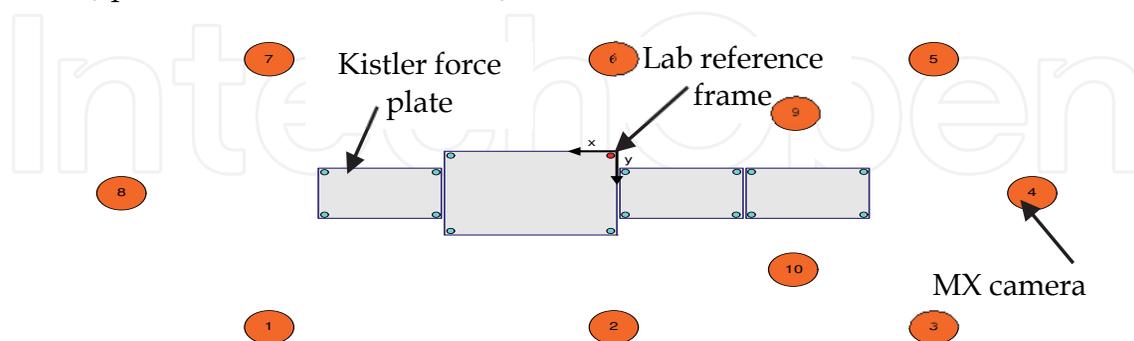


Fig. 3. Force plates and cameras configurations.

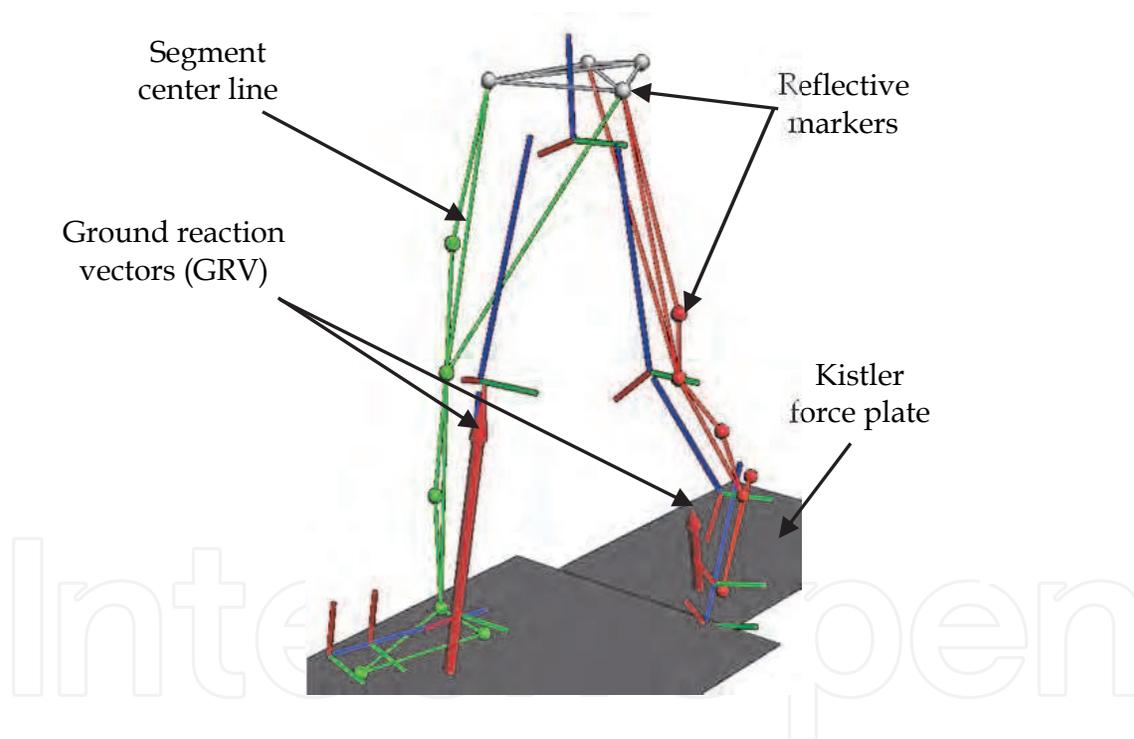


Fig. 4. Lower extremity stick figure and ground reaction vectors with markers and reconstructed segment's center lines.

2.1.3 Model construction

Once the gait session data was reconstructed and all gaps were filled then the C3D file that contains both the X, Y, and Z coordinates of the different markers (those markers was referred to later on as the grey markers) and the force plate data was transferred to the 3D lower extremity modelling phase. The model includes seven segments (Figure 5) that are scalable to the subject's dimensions; with 27 muscles in each leg been included, the exact

origin and insertion points as well as the different muscle parameters are all added in the model as well. Some artificial markers (will be referred to as the black markers) are introduced at the same landmarks as the ones collected in the Mocap session (Figure 6). Those black markers are driven by the grey markers and the sum of the error squared between the two at each time step was minimized according to the approach by Anderson et.al. (2006).

As shown in figure 5 and 6, by making the black marker as close as possible to the grey marker at all-time steps of the whole gait session it is possible to move the whole system of bones as well as ligaments and muscles attached to those bones in a smooth motion that is as close as possible to the captured motion of the patient. This guarantee that the segments will remain attached together, and the smooth normal motion of the whole system can be achieved which in turn facilitates the accomplishment of the kinematic analysis. After solving for the kinematics problem and the model is deemed to be functioning properly, the muscles' recruitment problem was solved next. This part of the model was trying to mimic the central nervous system (CNS) in recruiting the necessary muscles at the different time stamps during the gait session. Although there are many definitions and indices of muscle fatigue in the literature (Eltoukhy and Asfour, 2009), yet the way selected to approach such a problem was to adopt the idea utilized in human body which is based on minimizing the fatigue (equal to maximizing the endurance), which was done by utilizing the Min/Max criteria, such objective function was introduced in such problem by Rasmussen et al. (2001).

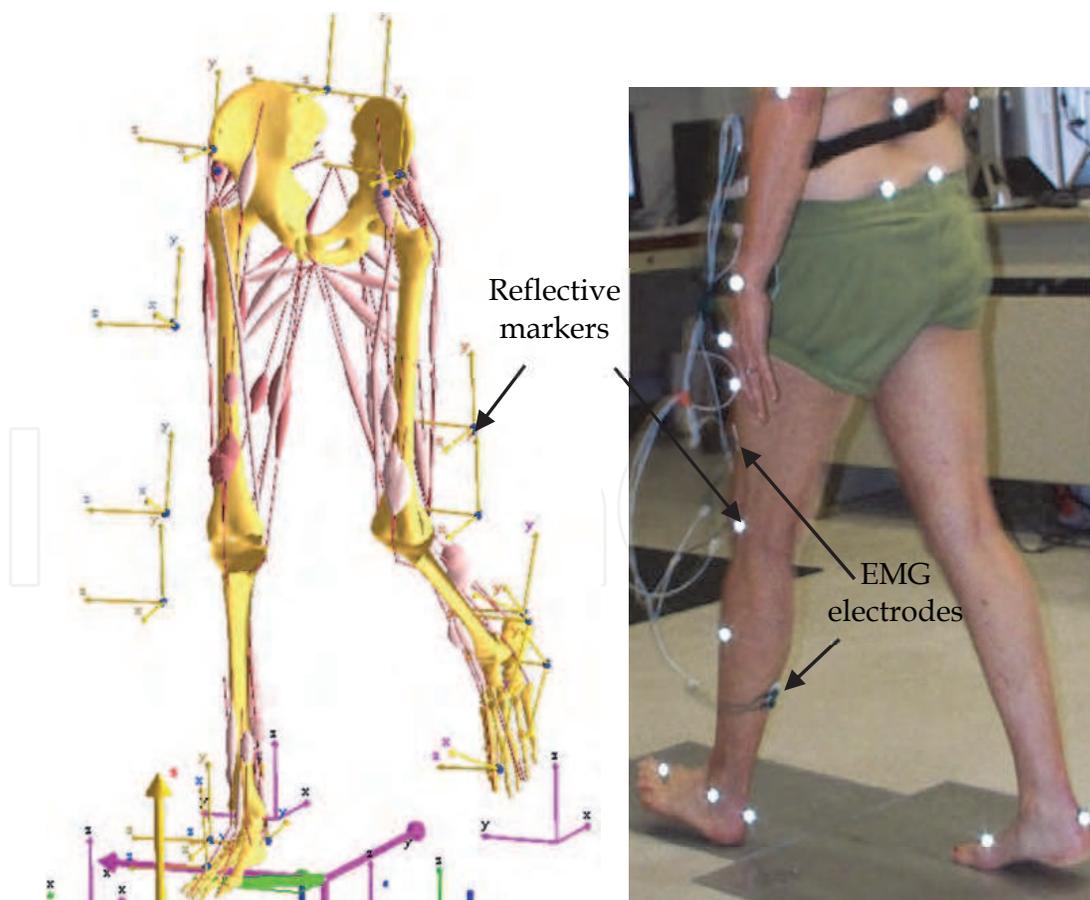


Fig. 5. 3D model of the lower extremity with muscles included and actual subject with EMG electrodes attached.

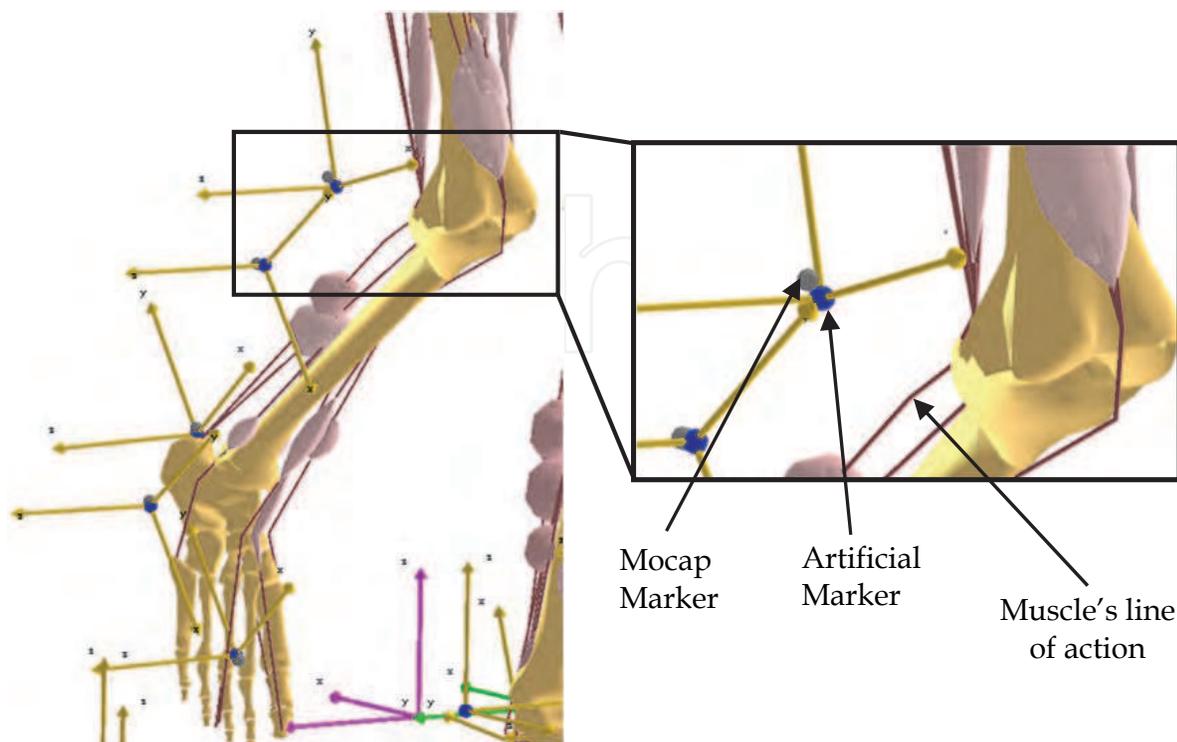


Fig. 6. Lower extremity model with bones and muscles included, each grey marker is followed by a black artificial marker at the same landmark. The sum of error squared between each pair of black and grey markers is minimized at all-time steps.

In their paper, the introduced criteria was applied to a 2D upper limb model composed of only two segments (upper arm and lower arm), three muscles, and one joint (elbow). The motion studied was a simulated motion of lifting a dumbbell and it was not an actual motion capturing experiment data. On the other hand, the developed model adds to that work by utilizing such powerful environment, AnyBody® technology, and provides a 3D model of all right and left legs and a pelvis, the model is driven by actual motion capturing data instead of a simulated (predetermined) motion. Also, the developed model is capable of simultaneously solve the optimization problem of recruiting and disrecruiting of all 54 muscles in both legs during such a complex motion like the gait.

During the muscle recruitment process the solver is trying to reach an optimum solution at each time step where both, the motion needed is achieved and the muscles' fatigue is postponed as much as possible.

A flow chart summarizing the different phases of the developed model is shown in figure 7. The main steps of the developed model can be described as follows:

- Measurement of subject's anthropometric data,
- Motion capturing (Mocap) data collection,
- Noise reduction obtained by a developed in-house LabVIEW signal processing module,
- Data reconstruction and labelling,
- Kinematic analysis utilizing an optimization algorithm by Anderson et al (2006). that minimizes the sum of error squared between any pair of markers (in this study, it's the actual and the artificial markers at the different land marks),

- Muscle recruitment using the capability of AnyBody® modelling environment, where it was possible to simultaneously optimize the recruiting and disrecruiting of 54 muscles in both legs across the whole gait cycle.

The model has a total of 7 bones which resulted in 42 degrees of freedom (d.o.f). On the other hand, only 24 d.o.f. were produced when the joints constraint were put in place, which means that a total of 18 marker coordinates were required to achieve the equilibrium (Table 1). In other words, there are more coordinates in the system than the d.o.f. Therefore, a subset of these coordinates is picked to kinematically drive the model.

Segment	# of Bones	Dofs Per Bone	Total
Foot	2	6	12
Shank	2	6	12
Thigh	2	6	12
Pelvis	1	6	6
			42

Segment	Joint Type	# of Bones	Dofs Per Bone	Total
Ankle	Universal	2	4	8
Knee	Hinge	2	5	10
Hip	Spherical	2	3	6
				24
Number of required marker coordinates				18

Table 1. Model components and degrees of freedom

The human body is composed of more than 200 bones bound and surrounded by soft tissues. To understand such a complex system, certain assumptions, simplifications and approximations must be made.

For instance, the foot is modelled as a single rigid body. The reason for this simplification is due to the complexity of foot anatomy, which is difficult to model mathematically. There are several other assumptions:

- The joint surfaces are frictionless.
- The mass of each segment is concentrated at its centre of mass.
- The estimates of body segment parameters provided by cadaver studies are sufficiently close to the real anthropometry of the person whose gait is being analysed. In practice, the errors from inaccurate segment parameters are thought to be very small, at least in normal gait (Pearsall & Costigan 1999).
- There is no co-contraction of the opposing agonist and antagonist muscles. If this occurs, a net moment will be calculated, which will be the difference between the moments generated by the two muscles.

The following section will describe the details of the kinematic analysis and the muscle recruitment problem. In summary, the model implemented is based on the method of inverse dynamics to determine the muscle forces from the kinematics data, while the muscle recruitment is utilizing the concept of minimizing the maximum muscle activity (Rasmussen et al, 2001), the models' drivers are developed based on the approach by Anderson et al. (2006) of optimizing the captured markers using the ViconNexus® motion capturing system.

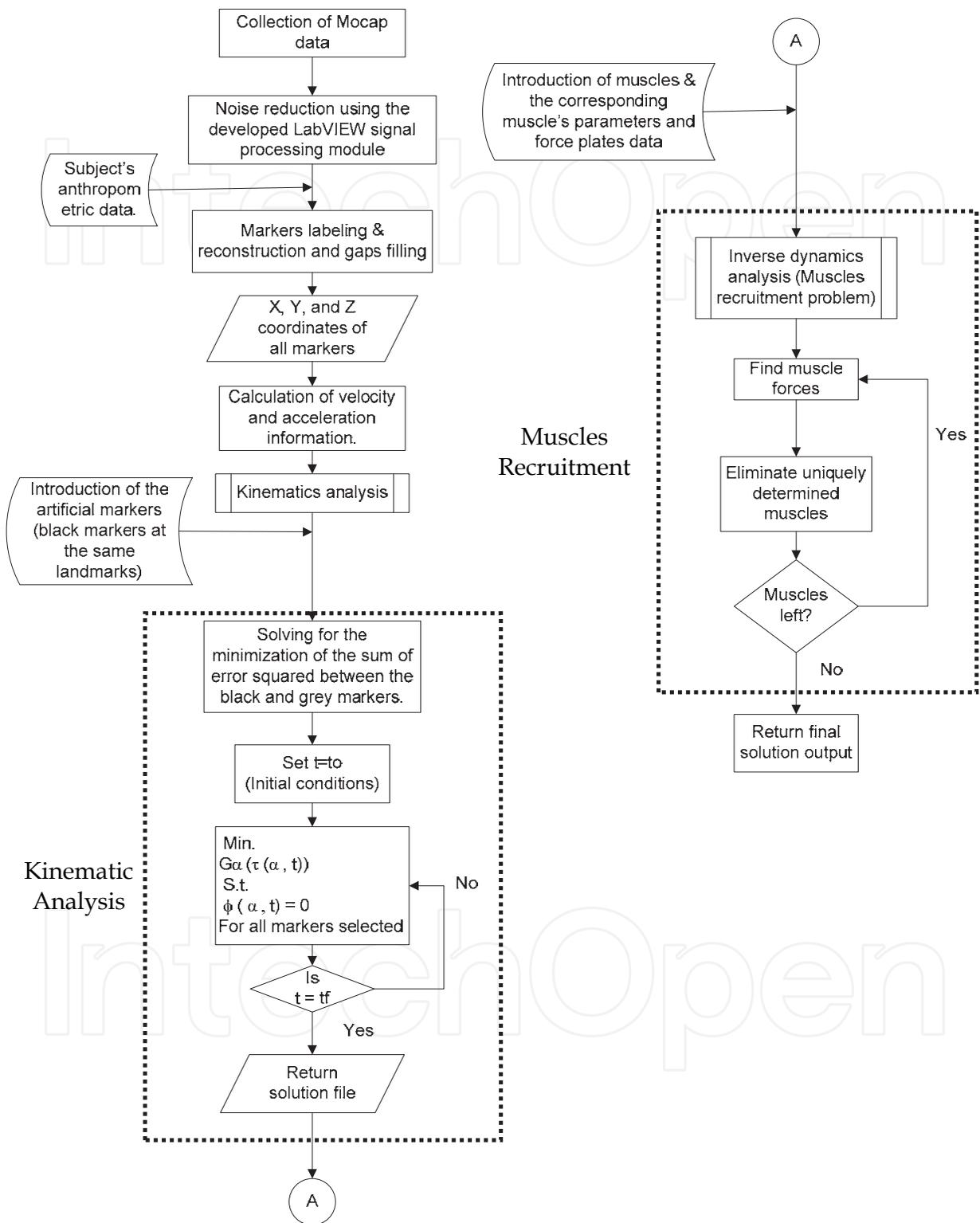


Fig. 7. Flowchart of the modelling steps and phases involved in the analysis of human gait cycle.

2.1.3.1 Kinematic analysis

When developing a musculoskeletal model from motion capture data for inverse dynamics analysis, the resulting set of equations usually will be over-determinate, i.e. the measurement results in more measured degrees-of-freedom than the ones of the model. Several solutions to this problem have been posed in the literature.

They are essentially split into two groups: 1) methods that work on a segment-to-segment basis, and 2) methods that use model information, such as joint constraints, to reduce the effects of skin-artefacts. A general solution to the problem of kinematic over-determinacy when driving a mechanical model from measured data in the form of marker trajectories is implemented in this study.

The solution suggested by Anderson et al. (2006) is based on solving an appropriate constrained weighted least-squares optimization problem for each discrete time step with the system coordinates as unknowns.

The first step in inverse dynamics is always to perform kinematic analysis to find the positions, velocities and acceleration of the time-dependent system coordinates, $\alpha(t) \in \alpha$, i.e. given some system description it is desired to find $\alpha(t)$, $\alpha'(t)$ and $\alpha''(t)$.

To accommodate this over-determinacy, it is assumed that it is possible to split the position analysis equation into two sets:

$$\chi(\alpha, t) = \begin{bmatrix} \tau & (\alpha, t) \\ \phi & (\alpha, t) \end{bmatrix} \quad (1)$$

Where $\tau \equiv \tau(\alpha, t)$ is a set of equations that only have to be solved “as well as possible” in some sense and the remaining $\phi \equiv \phi(\alpha, t)$ equations have to be fulfilled exactly.

In this study, where the musculoskeletal model was derived from motion capture data, the choice of these sets was that the experimental data belongs to τ and joint constraints and additional driver equations to ϕ .

In other words, it is required to solve the following optimization problem:

$$\text{Min } G\alpha(\tau(\alpha, t)) \quad (2)$$

$$\text{s.t. } \phi(\alpha, t) = 0 \quad (3)$$

This is done by automatically taking all the marker coordinates into account and then taking the minimized deviation between the markers in the model and the markers captured in the motion capturing experiment (Figure 6), i.e., minimizing the square sum of errors. This process is done in an offline mode, where the data is analyzed and then fed back to the model.

Having solved the optimization problem, the system coordinates, α , will be known for the discrete time steps where the optimization problem is solved. However, it still remains to find the velocities and accelerations which are accomplished by differentiating the position data to determine the velocities and a second time to determine the accelerations.

2.1.3.2 Muscle recruitment

The second part of the model is the muscles recruitment problem, where in which the model mimics the CNS in its coordination of muscles during complex activities such as human locomotion.

The basic optimality assumption considered in the model is that the body attempts to use its muscles in such a way that fatigue is postponed as far as possible. Which leads to the idea of minimizing maximum muscle activity (Rasmussen et al., 2001). Thus, the model will recruit muscles according to the following criterion, put in the general form:

Minimize:

Maximum muscle activity

Subject to:

- Equilibrium equations fulfilled.
- Muscles can pull only.

It was also proven by Rasmussen et al. (2001) that the min/max criteria is identical to both soft saturation and polynomial objective functions at high power. And due to the advantages that the min/max possesses, it was decided to apply it as the recruitment criteria.

The effect of implementing the min/max criteria in the muscle recruitment problem is that they tend to form groups; meaning that if there are a number of muscles crossing the same joint and have the same strength; they will form a group in the recruitment. This is because the min/max criteria tries to decrease the activity in all muscles simultaneously, which guarantees that there is no other muscle recruitment pattern that could lead to smaller muscle activity.

2.1.3.3 Noise reduction

The model interpolates the data by a smooth spline interpolation, and it is possible to suppress the noise to some extent by increasing the order of the spline interpolation and/or by down sampling the data. Yet, the best way to minimize the noise is to low pass filter the data before sending them to the model. This will get rid of most of the high frequency noise in the motion capture data, this goal was achieved by developing another module based on LabVIEW programming scheme.

The effect of filtering the data using the developed signal processing module is that it dramatically reduces the time needed by the solver to find the optimum solution of the optimization problem during the kinematic analysis stage (Figure 8). The vertical axis in the

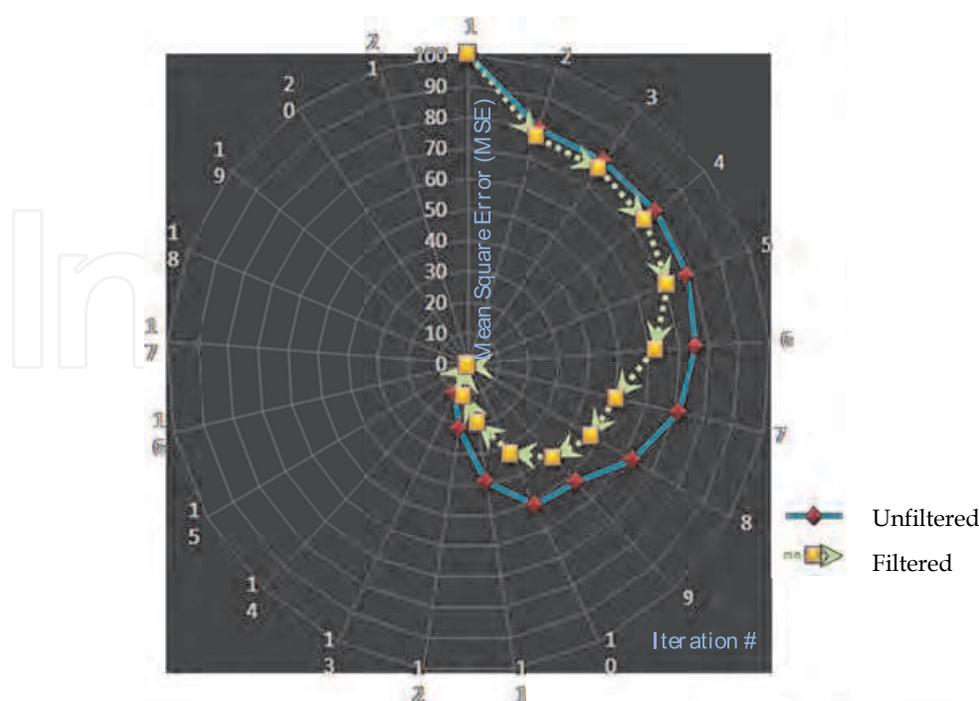


Fig. 8. Mean square error plots of both filtered and unfiltered data

figure depicts the mean squared error (MSE) resulted at each iteration of the solution process of the objective function. The outer circle is the iteration number of the corresponding MSE value.

As shown in figure 8, the dotted curve represents the filtered data indicates that the solver reaches an optimum solution at a fewer number of iterations as compared to the unfiltered data (solid curve), where more iterations were needed to reach an optimum solution for the objective function.

Once the model is validated against measured data it can then be used to analyze the effect of muscles on a hip or knee prosthesis, especially when considering different surgical techniques.

3. Results and discussion

The following section will introduce some of the model outputs obtained in comparison to one of the literature resources available which is the Hip98 data base (Bergmann et al. 2001). All data shown were normalized by the subject's weight. Figure 9 shows a sample of the knee reaction forces obtained from one of the patients studied in the laboratory as compared to the pattern obtained by Harrington (1992).

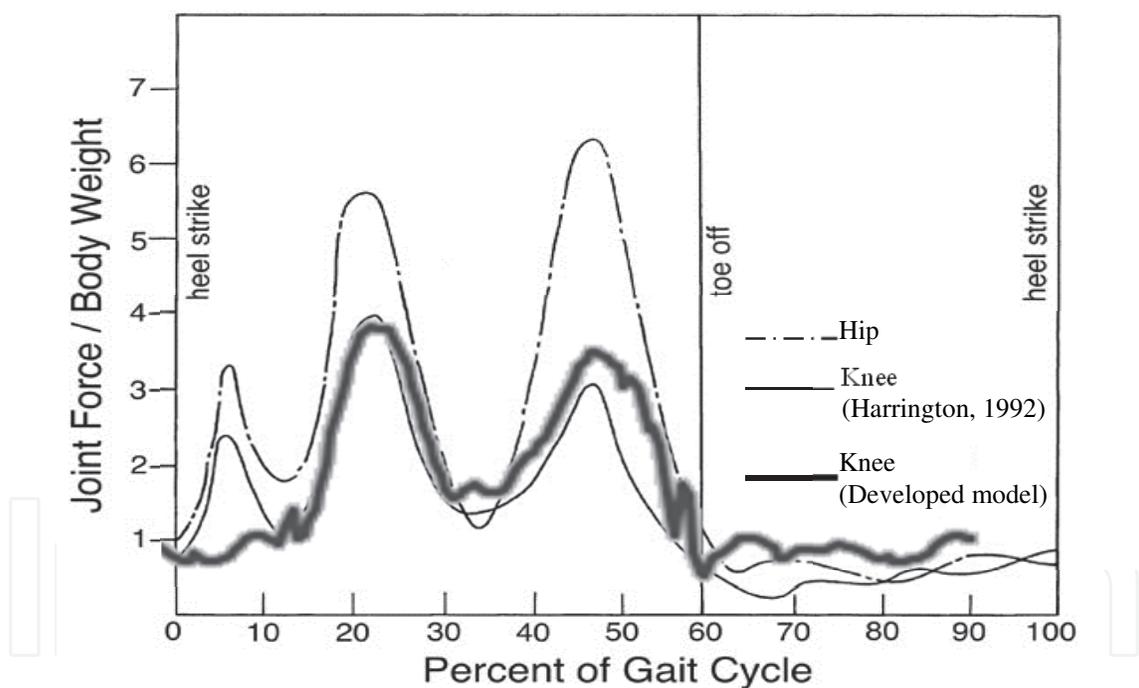


Fig. 9. Knee reaction forces pattern. (Harrington, 1992)

The moments were then calculated. Figure 10 (A, B, and C) shows the hip moments generated in the flexion / extension, internal / external rotation, and abduction / adduction directions respectively.

The force and moment patterns and magnitude predicted were in a great agreement with the measured values in the literature (Hip98). The ground reaction forces utilized in the model were compared to the forces reported in the Hip98 data base as shown in figure 11 (A, B, and C). The figure depicts the ground reaction forces in the inferior, lateral, and anterior directions respectively.

3.1 Case study: Total knee replacement

A total of fifteen subjects participated in this study, five subjects in each group. Based on the statistical analysis conducted it was evident that the allograft group had less significant gait deviations than did those in the metallic group. In particular, the metallic group patients had greater gait deviations both in the loading-response phase, and in the Fore-Aft terminal stance phase. In conclusion, our findings suggest that implantation of hinged total knee prosthesis with removal of one or two of the quadriceps muscles provide good functional results during gait. Additional studies should be performed to evaluate the relationship between the number and extent of the quadriceps heads excised and both the knee mechanics during gait and the long-term survival of the prosthesis.

The comparison between the gait results obtained using the musculoskeletal model and a very relevant experimental research conducted by Benedetti et al. (2000), it was also concluded that the introduced model can provided a high confidence level if used in applications that involves joint replacements which ultimately can be utilized in surgery planning.

3.1.1 Subjects and procedures

Demographics of the Control Group

Five subjects with no history or complaints of known walking problems, knee injuries and postural instability volunteered for this experiment.

Demographics of the Patient Group

Ten patient subjects was the maximum number that was available to participate in this study. All patient subjects had distal femoral knee replacement surgery. Five subjects were allograft patients while the other five were metallic patients. The Following table summarizes the demographics of all three subject groups.

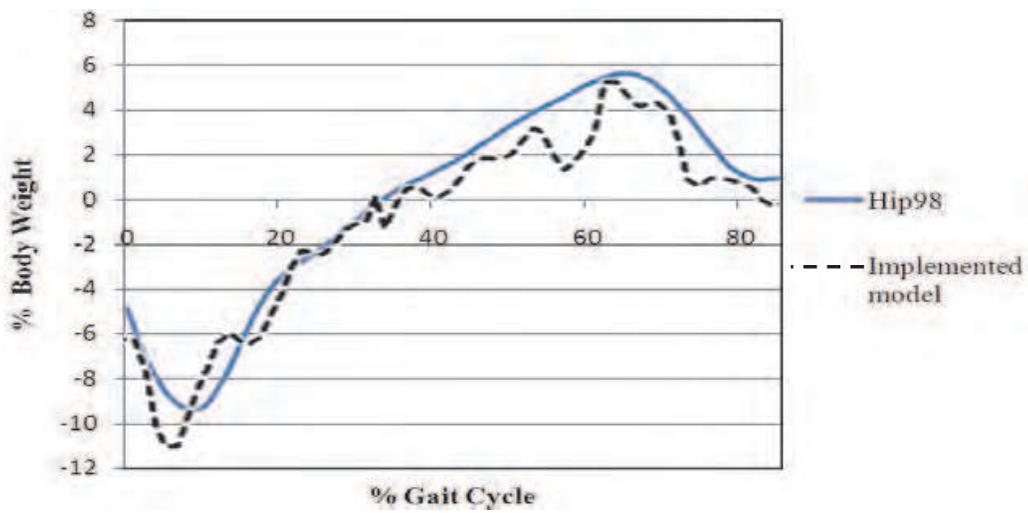
The procedures performed on the study subjects were wide local excision of the tumor and reconstruction using either a Stryker (tm) MRS rotating hinged knee or an osteoarticular allograft. The MRS knee was placed after radical resection of the distal femur. The distal femur was prepared by reaming the intramedullary canal with the appropriate sized reamer to achieve the greatest diameter stem with an appropriate cement mantle (2mm circumferential). The tibia was prepared with standard cutting jigs.

	Control	Metallic	Allograft
No of Subjects	5	5	5
Age (years)	26.6 ± 0.9	42.3 ± 5.9	28.6 ± 5.6
Height (cm)	175.8 ± 1.9	175.8 ± 8.5	163.0 ± 8.0
Weight (kg)	71.8 ± 3.9	89.6 ± 15.5	66.0 ± 8.3

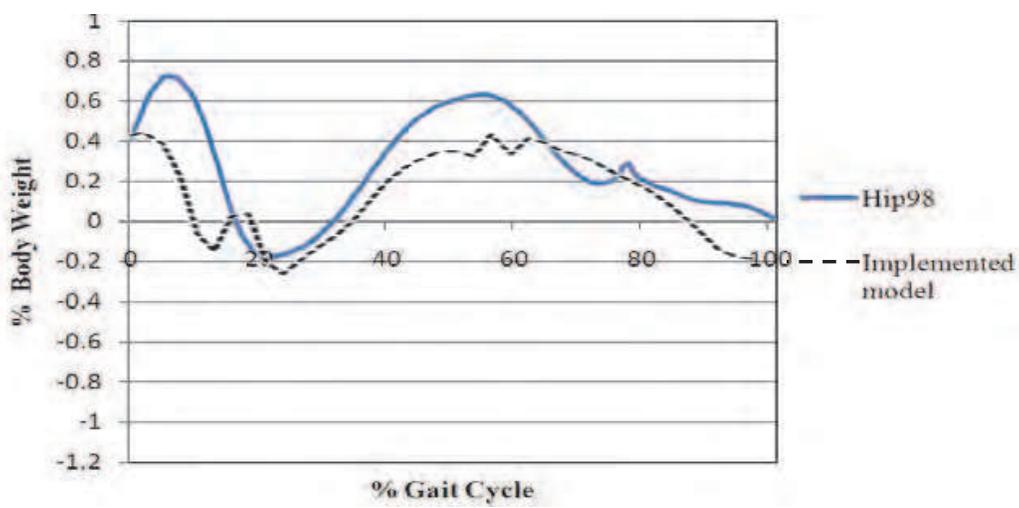
Table 2. Demographics of Subjects

The tibial and femoral canals were filled with cement after pulse irrigation and drying using modern cement techniques. The components were segmental and prepared on the back table to reconstruct the longitudinal dimensions of the defect. The components were assembled using inner bearings an axle and rotating metal hinge. In some cases, a gastrocnemius flap was turned up into the defect to fill the dead space and to help tether the lateral pull of the extensor mechanism. The allografts were prepared differently.

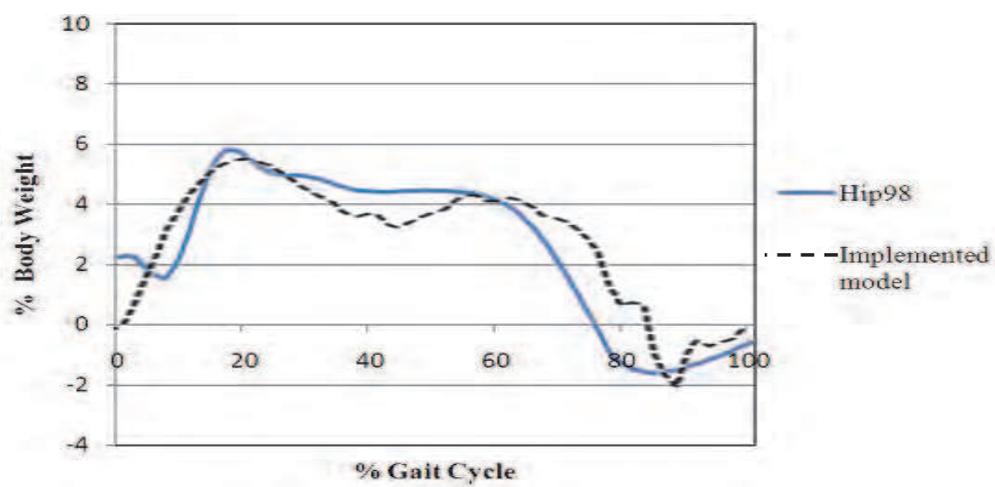
The patient's menisci and portions of the lateral, medial collateral ligaments along with the anterior and posterior cruciate ligaments were retained. The allograft was aseptically



(a)



(b)



(c)

Fig. 10. Hip moments generated in the following directions: (A) Flexion/Extension, (B) Internal /External rotational, and (C) Abduction/Adduction.

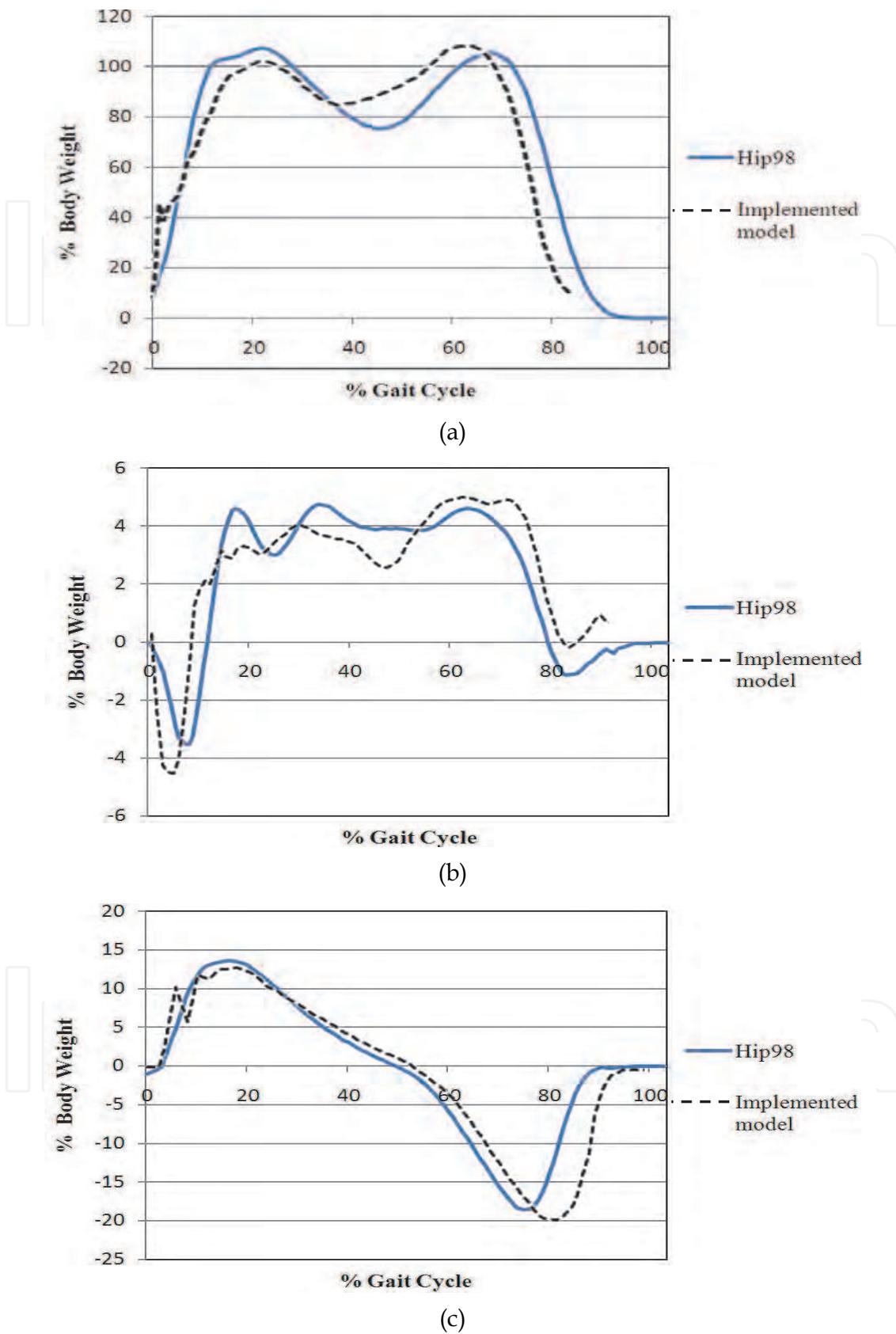


Fig. 11. Ground reaction forces in the following directions: (A) Inferior, (B) Lateral, and (C) Anterior.

harvested from a suitable donor; the cartilage was preserved with DMSO and frozen in a controlled fashion to -80 degrees for storage. In the operating theatre, this preselected graft was then thawed in lactated ringer solution and cut on the back table to the appropriate length. A plate or plates were then affixed to the graft and the soft tissue structures were then reconstructed. The reconstruction began with the posterior capsule followed by the posterior cruciate, the lateral collateral ligaments and finally the anterior cruciate ligament. The meniscomfemoral attachments to the allograft were repaired following this. The graft itself was then anchored to the patient's host bone with plates and screws.

The participants were given instructions including an explanation of the test procedures, proper attire, and the expected duration of the testing. The data was collected at the University of Miami Biomechanics Research Laboratory, USA, with an approval from the Internal Review Board (IRB). 48 reflective markers were placed at the different body landmarks (e.g. joints center lines and segments). First, a static trial was conducted where the patient stands at a tee pose which is then used for markers labeling. Then five dynamics trials were performed. Each subject was instructed to walk at his normal speed across the laboratory and on top of the four force plates.

3.1.2 Instrumentation and data collection

The laboratory incorporates a ViconNexus® Motion Capturing System (Oxford Metrics, United Kingdom). The motion capturing system integrates and synchronizes four Kistler force plates (Model: 9253B), ten MX cameras, two high speed reference video cameras, and wireless Noraxon EMG system. The MX cameras provide 1024 x 1024 pixel resolution and frame rates up to 250 Hz.

3.1.3 3D Lower extremity musculoskeletal model

Once the motion capturing data was reconstructed and all gaps were filled then the C3D file, which contains the X, Y, and Z coordinates of the different markers as well as the force plate data, is imported to the 3D lower extremity modeling phase. The three-dimensional musculoskeletal model of the lower extremity introduced by Eltoukhy and Asfour (2009) was utilized in this study.

The model consisted of seven rigid body segments connected with frictionless joints, these segments are the pelvis, right and left thighs, right and left shanks and right and left feet. Hip joints were modeled as universal joints that have three degrees of freedom. Knee joints were modeled as hinge joints, while ankle joints were modeled as biaxial joints. Hip joint flexion/extension, adduction/abduction and internal/external rotation as well as ankle joint dorsi/plantar flexion and inversion/eversion were all allowed (Figure 12). The model is scalable to the subject's dimensions; with 27 muscles in each leg been included, the exact origin and insertion points as well as the different muscle parameters are all added in the model as well.

4. Statistical analysis

Among the different gait variables measured, a specific subset of these variables was carefully selected. These variables were then used in the statistical analysis. Figure 13 shows the ground reaction forces in the X, Y, and Z directions as well as the knee flexion angle measured during the gait session. The figure also depicts the specific force and angle values at the different point of times during the gait cycle that were analyzed in the statistical analysis phase.

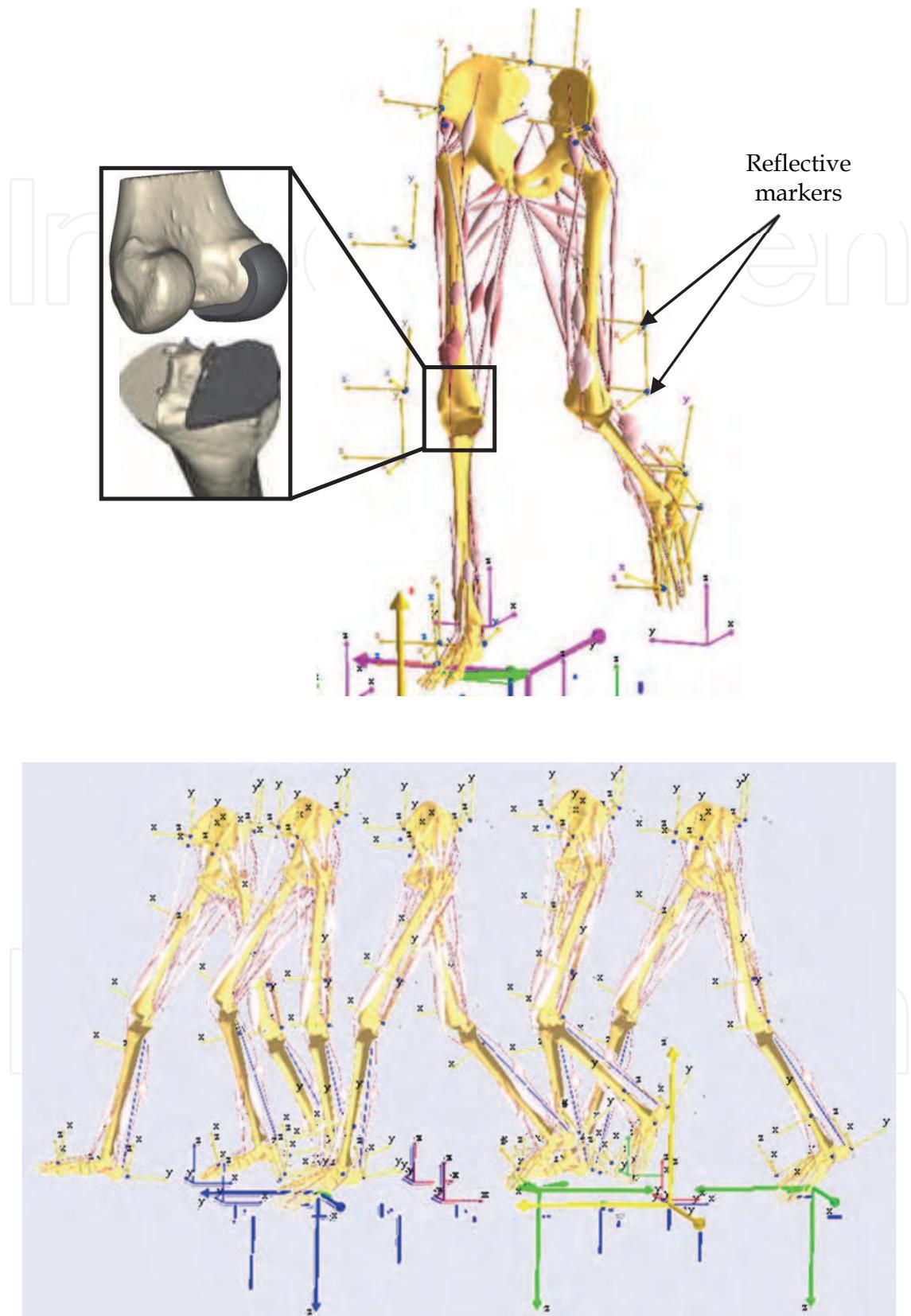
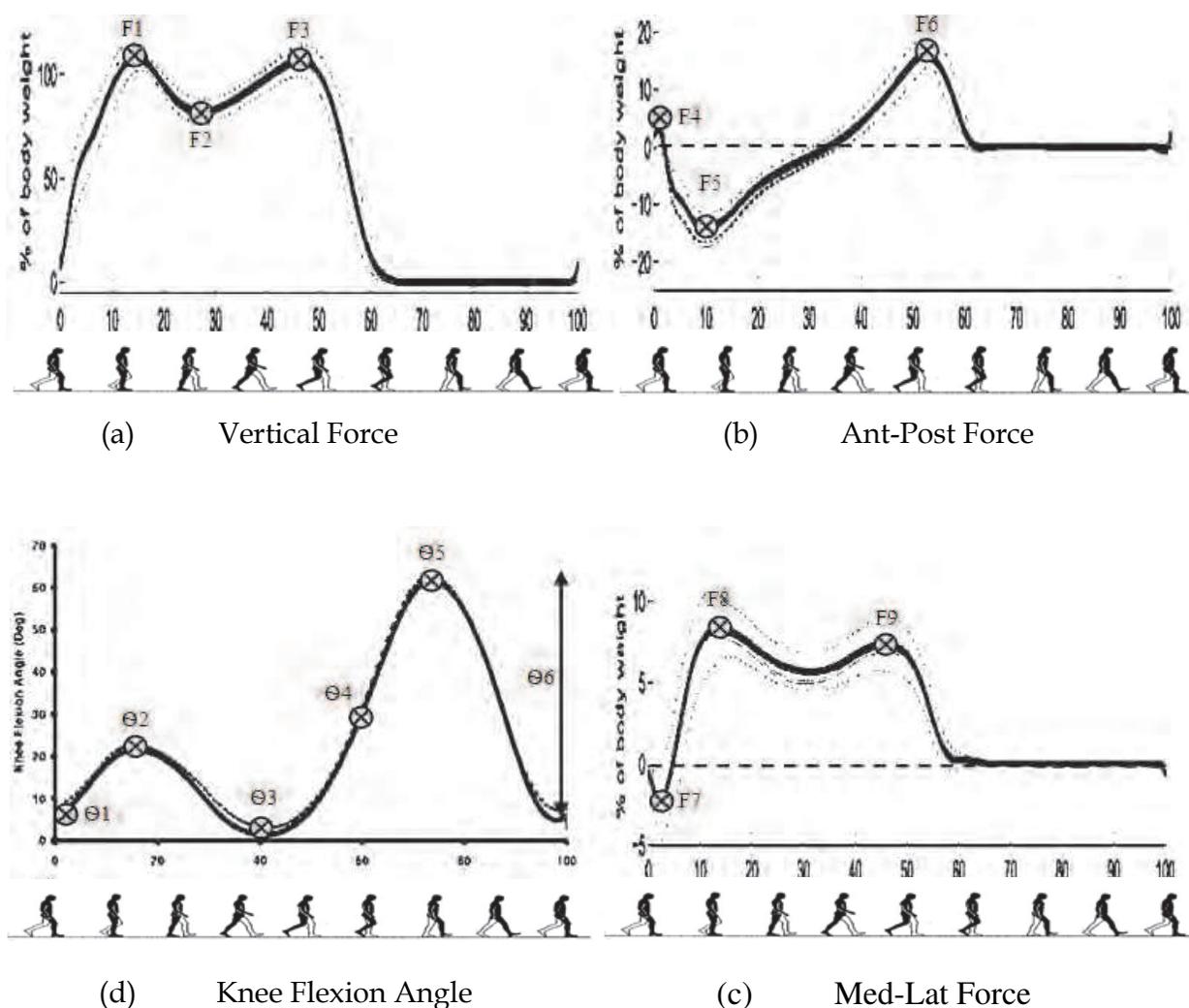


Fig. 12. Modified lower extremity model introduced by Eltoukhy and Asfour, 2009.



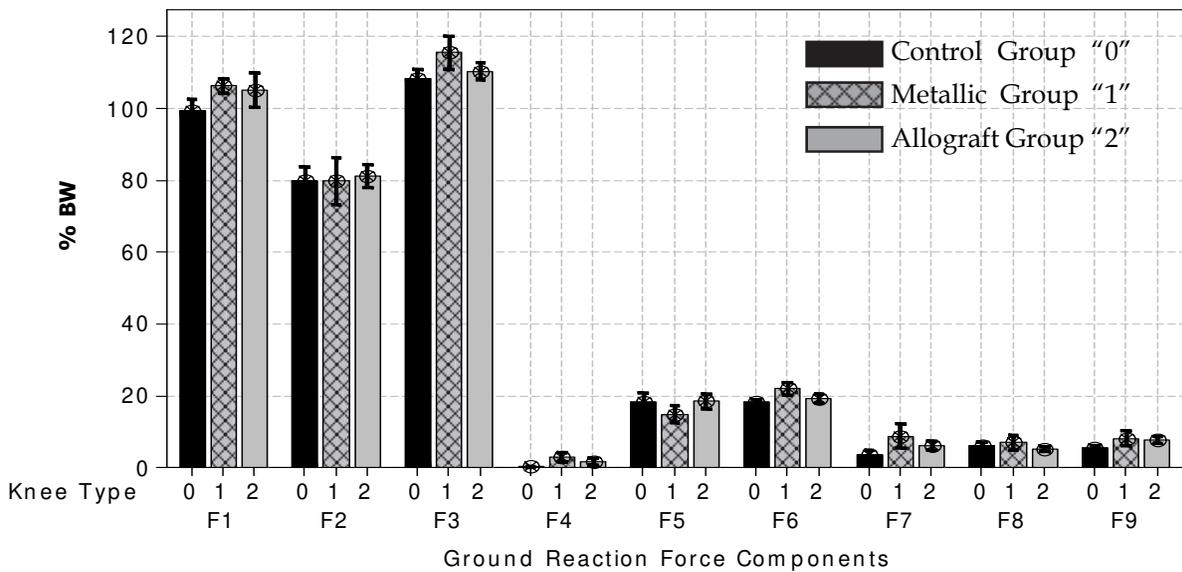
F1:Maximum vertical loading response	F6:Maximum Fore-Aft terminal stance	Θ2:Maximum Flexion loading response
F2:Maximum vertical midstance	F7:Maximum Med-Lat loading response	Θ3:Maximum Extension stance
F3:Maximum vertical terminal stance	F8:Maximum Med-Lat midstance	Θ4:Flexion toe-off
F4:Maximum Fore-Aft loading response	F9 :Maximum Med-Lat terminal stance	Θ5:Maximum Flexion swing
F5:Maximum Fore-Aft midstance	Θ1:Flexion heel-strike	Θ6:Total sagittal plane excursion

Fig. 13. The gait parameters measured.

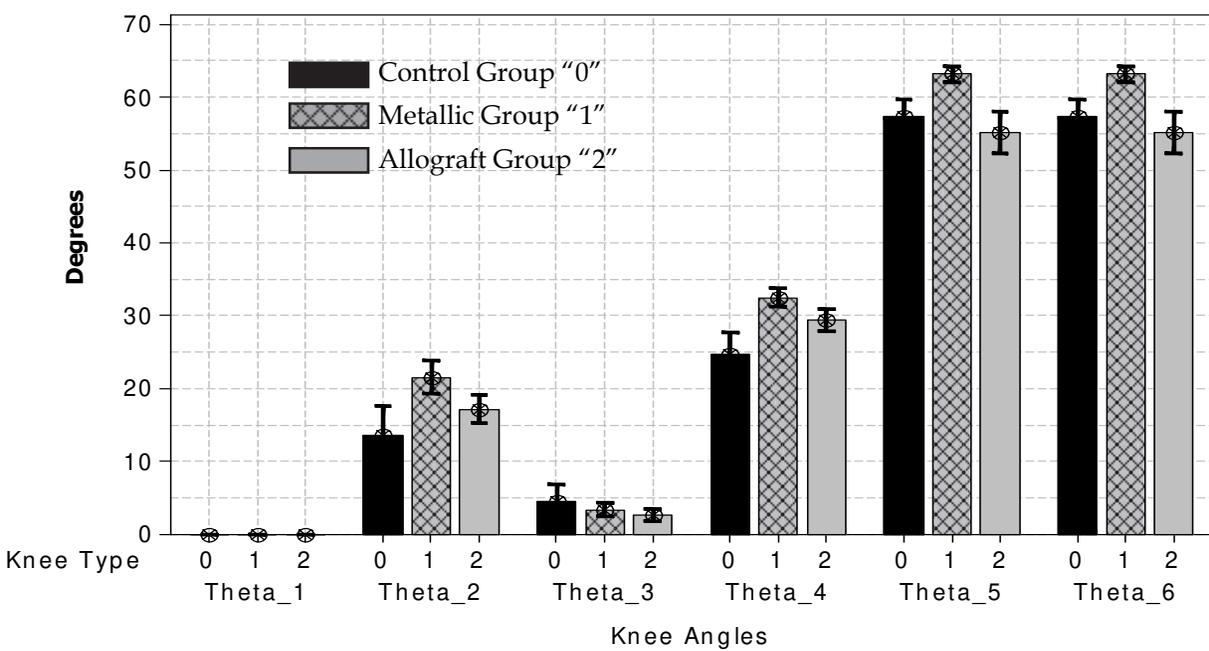
As shown in the figure, three main points (F1, F2, and F3) in the vertical force component (Figure 6-a); also, F4, F5, and F6 were recorded from the Ant-Post force component (Figure

6-b). The last three force components recorded were the F7, F8, and F9 and that is from the Med-Lat force plot (Figure 6-c). Six angle values were recorded from the knee flexion angle component. These angle components were Θ_1 to Θ_6 (Figure 6-d).

The data collected from the two patient groups (metallic and allograft knee joints) was plotted along with the control group (normal knee joint). Figure 14 depicts two interval



(a)



(b)

Fig. 14. Interval plots of (a) the ground reaction forces (F1 to F9) and (b) the knee flexion angle (Θ_1 to Θ_6) for all three groups, Control "0", Metallic "1", and Allograft "2".

plots for both the ground reaction forces as well as the knee flexion angle measured. As shown in figure 14, each of the force and angle components was plotted in the form of the mean and standard deviation values for all three groups, control, metallic, and allograft.

For the vertical ground reaction force (Z direction) components, both the metallic and allograft groups showed higher average forces when compared to the control group, for the maximum vertical loading response, F1, and the maximum vertical terminal stance, F3 while they showed almost an equal magnitude for the maximum vertical midstance, F2.

The Ant-Post force values (F4-F6) resulted in a similar pattern as the vertical forces where the metallic and allograft groups showed higher values when compared to the control group for both F4 and F6, yet, in case of the maximum Fore-Aft midstance force, F5, the metallic group resulted in lower magnitude while the allograft group resulted in higher forces when compared to the control group. Similar pattern was also obtained in case of the Med-Lat force values, only the F8, maximum Med-Lat midstance, showed higher forces for the metallic group and similar mean value for allograft group when compared to the control group. In regard to the flexion knee angle values, the following conclusions can be drawn. In general, the metallic group showed higher knee flexion angle values ($\Theta 2$: Maximum Flexion loading response, $\Theta 3$: Maximum Extension stance, $\Theta 4$: Flexion toe-off, $\Theta 5$: Maximum Flexion swing, and $\Theta 6$: Total sagittal plane excursion) as compared to the allograft group, in other words, the allograft group showed stiff-knee gait pattern. Note that only in case of the $\Theta 3$ (maximum extension stance), the control group showed higher mean value than the metallic and allograft groups which can be explained because of the range of motion loss resulted from the surgery when compared to the control group.

A statistical analysis of the gait data collected was performed as follows; two Multi-Analysis of Variance (MANOVA) were conducted for both the ground reaction force components as well as the knee joint angle. The factor tested was the knee type (Normal, Metallic, and Allograft), on the other hand the variables tested were the different force components of the three ground reaction forces in the X,Y, and Z directions (F1 to F9) as well as the knee angles ($\Theta 2$ to $\Theta 6$). The data was first tested for normality. The basic data descriptive statistics as well as the MANOVA results are summarized in table 3, 4, and 5 respectively.

Variable	Count	Mean \pm StDev	Variable	Count	Mean \pm StDev
F1	30	103.7 \pm 5.6	$\Theta 1$	30	0.0
F2	30	80.3 \pm 6.6	$\Theta 2$	30	17.5 \pm 5.1
F3	30	111.5 \pm 5.4	$\Theta 3$	30	3.6 \pm 2.1
F4	30	1.6 \pm 1.6	$\Theta 4$	30	28.9 \pm 4.3
F5	30	17.2 \pm 3.5	$\Theta 5$	30	58.5 \pm 4.6
F6	30	19.8 \pm 2.4	$\Theta 6$	30	58.5 \pm 4.6
F7	30	6.1 \pm 3.6			
F8	30	6.0 \pm 1.9			
F9	30	7.0 \pm 2.3			

Table 3. Ground reaction forces and knee angles descriptive statistics

Criterion	Test Statistic	F	DF		P
			Num	Denom	
Wilks'	0.04444	7.903	18	38	0.000
Lawley-Hotelling	7.87640	7.876	18	36	0.000
Pillai's	1.56107	7.904	18	40	0.000
Roy's	5.31098				

s = 2 m = 3.0 n = 8.5

Table 4. MANOVA for Knee Replacement Type (Ground Reaction Forces)

Criterion	Test Statistic	F	DF		P
			Num	Denom	
Wilks'	0.05324	20.003	8	48	0.000
Lawley-Hotelling	9.14198	26.283	8	46	0.000
Pillai's	1.40680	14.822	8	50	0.000
Roy's	8.07144				

s = 2 m = 0.5 n = 11.0

Table 5. MANOVA for Knee Replacement Type (Knee Angles)

As shown in the MANOVA tables and based on the Wilks criteria, the data suggests that the type of knee replacement is associated with changes in gait pattern. As tables 4 and 5 indicate, other similar tests, namely Pillai’s Trace, Hotelling’s Trace, and Roy’s Largest Root test, also lead to the same conclusion. The Wilks’ lambda test is used to measure the overall significance of the model. When the overall model is significant, then the significance of the individual variables can be pursued. After statistically significant evidence was obtained from the MANOVA tests, the individual ANOVA analyses for both the ground reaction forces (GRF) and knee flexion angles were performed and the results were summarized in table 6.

Table 6 basically shows the results of the individual ANOVA tests for all GRF components and knee flexion angles. The table also shows the P values, as well as the mean and standard deviations for all variables and that is for each group separately. The table also summarizes the post-hoc analysis results; the post-hoc performed compared the control vs. the metallic, the control vs. the allograft, and the metallic vs. the allograft groups. In general, the allograft showed similar values for both the forces and angles during gait when compared to the control group. In other words, no statistically significant differences were found between the two groups, that is true for all force values (except for F4 and F9) and knee angles (except for Θ_4). On the other hand, the metallic group showed statistical differences in most of the force values (except for F8) and knee angles (except for Θ_3) when compared to the control group. Finally, to exclude the age factor as a possible reason behind the finding that the allograft group resulted in less gait deviations compared to the metallic group in comparison to the control group, the following statistical analysis was conducted. First a separate MANOVA analysis was conducted with the “Age” set as the factor been modelled and the variable set were the flexion knee angles (Θ_2 to Θ_5) as shown in table 7.

Variable	P Value*	P-Value (Post-hoc analysis)					
		Control Group	Metallic Group	Allograft Group	Control Group Vs. Metallic Group	Control Group Vs. Allograft Group	Metallic Group Vs. Allograft Group
GRF		Mean±Std	Mean±Std	Mean±Std			
F1	0.010	99.5±4.2	106.3±3.0	105.2±6.8	Sig.	NS	NS
F2	NS	79.8±5.6	79.8±9.2	81.2±4.6	NS	NS	NS
F3	0.006	108.5±3.3	115.6±6.6	110.4±3.2	Sig.	NS	Sig.
F4	0.001	0.2±0.07	2.7±1.8	1.7±1.3	Sig.	Sig.	NS
F5	0.028	18.3±3.3	14.8±3.3	18.4±2.9	Sig.	NS	Sig.
F6	0.001	18.2±1.1	21.9±2.5	19.3±1.8	Sig.	NS	NS
F7	0.004	3.6±1.7	8.7±4.6	6.0±1.9	Sig.	NS	NS
F8	NS	6.2±1.1	6.9±2.7	5.0±0.9	NS	NS	NS
F9	0.007	5.3±1.0	8.0±2.9	7.8±1.3	Sig.	Sig.	Sig.
Knee angles							
θ 2	0.001	13.6±5.6	21.6±3.2	17.2±2.7	Sig.	NS	NS
θ 3	NS	4.6±3.1	3.4±1.3	2.7±1.3	NS	NS	NS
θ 4	0.000	24.7±4.3	32.5±1.8	29.5±2.1	Sig.	Sig.	NS
θ 5	0.000	57.3±3.2	63.2±1.5	55.1±4.0	Sig.	NS	Sig.
θ 6	0.000	57.3±3.2	63.2±1.5	55.1±4.0	Sig.	NS	Sig.

* According to analysis of variance unless otherwise indicated. Sig.: significant. NS: not significant.

Table 6. Descriptive Statistics and ANOVA results

Criterion	Test Statistic	F	DF		P
			Num	Denom	
Wilks'	0.00016	1.499	56	9	0.265
Lawley-Hotelling	75.73244	0.676	56	2	0.763
Pillai's	3.16755	1.359	56	20	0.227
Roy's	61.09191				

s = 4 m = 4.5 n = 0.0

Table 7. MANOVA for patients' age (Knee Angles)

As shown in table 7 and based on the three MANOVA indices, there was a clear evidence that no statistically significant effect of the patients' age on the knee kinematics. The second analysis performed was a set of three MANOVA tests of the ground reaction force components (F1 to F9) with the "Age" set as the factor tested. Table 8 summarizes the

MANOVA outputs for these three tests. It was also evident that in case of the kinetics, the patients' age had failed to show any statistically significant effect on the ground reaction force components produced during gait.

Vertical Forces (F1-F3)		DF			
Criterion	Test Statistic	F	Num	Denom	P
Wilks'	0.00209	1.613	42	9	0.227
Lawley-Hotelling	24.80947	0.985	42	5	0.579
Pillai's	2.53192	1.932	42	15	0.084
Roy's	13.96643				
Fore-Aft Forces (F4-F6)		DF			
Criterion	Test Statistic	F	Num	Denom	P
Wilks'	0.00266	1.467	42	9	0.279
Lawley-Hotelling	29.76959	1.181	42	5	0.475
Pillai's	2.39169	1.404	42	15	0.242
Roy's	22.86074				
Med-Lat Forces (F7-F9)		DF			
Criterion	Test Statistic	F	Num	Denom	P
Wilks'	0.00625	20.003	42	9	0.513
Lawley-Hotelling	16.62683	0.660	42	5	0.794
Pillai's	2.33854	1.263	42	15	0.321
Roy's	10.82468				

Table 8. MANOVA for patients' age (Ground Reaction Forces)

5. Conclusions

The goal of this research work shown in the chapter was to introduce the whole process of developing and validating a 3D lower extremity musculoskeletal model and to test the ability of the model to predict the muscles recruitment of the different muscles involved in human locomotion as well as determining the corresponding forces and moments generated around the different joints in the lower extremity. Therefore the model can be applied in one of the important fields of orthopaedics which is joint replacement; the case study used in such application is the total knee replacement. The knee reaction forces were compared to the pattern obtained by Harrington (1992), where the hip moment components (Flexion/extension, internal/external, and abduction/adduction) were all compared to the patterns obtained from the Hip98 data base.

As it was shown in the different forces and moments graphs, the model was able to produce very close results when comparing pattern and magnitude to the literature data. Thus, this 3D biomechanical model is sophisticated enough to be used for surgery evaluation such as in total knee replacement, where the damaged cartilage and bone are removed from the surface of the knee joint and replaced with a man-made surface of metal and plastic.

The three-dimensional (3D) musculoskeletal model of the lower extremity introduced by Eltoukhy and Asfour (2009) has been utilized in the joint replacement applications. The objective of this case study was to test the applicability of the introduced model in situations

that involves joint replacement and to show that the model can be used for such applications such as surgery planning. The case study of the research work presented in this chapter involved the comparison of the gait pattern between two main knee joint types, Metallic and Allograft knee joints against normal subjects (Control group). A total of fifteen subjects participated in this study, five subjects in each group. Based on the results obtained from the MANOVA tests, the allograft group had less significant gait deviations than did those in the metallic group. In particular, the metallic group patients had greater gait deviations both in the loading-response phase, and in the Fore-Aft terminal stance phase. It was concluded that based on the study conducted and the statistical evidence obtained that the introduced model can be used for applications that involves joint surgeries such as knee replacement that ultimately can be utilized in surgery evaluation.

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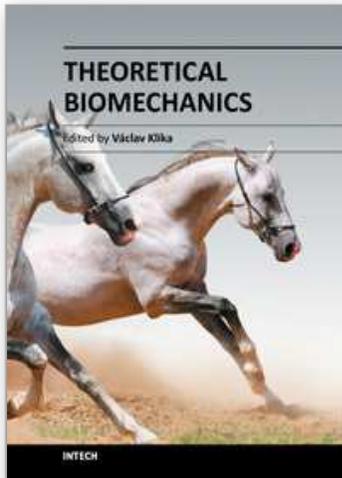
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