Musculoskeletal multibody simulations for the optimal tribological design of human protheses: the case of the ankle joint

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Abstract. A thorough determination of the loading of the ankle joint is useful both for the optimal design of prostheses and for the preclinical testing in terms of tribological performances. In vivo measurements of joint forces are usually not easy in the in-vivo settings, then non-invasive in-silico methods should be considered. Nowadays resultant joint loads can be reliably estimated by using musculoskeletal modelling in an inverse dynamic approach, starting by motion data obtained in gait analysis laboratories for several human activities. The main goal of this study was to provide a set of dynamical loading curves obtained by the AnyBody Modelling SystemTM (AMS) computer software starting from ground reaction forces and kinematic data obtained by Vaughan et al. in the case of human normal gait. The model accounts for 70 Hill modelled muscles and the muscular recruitment strategy was choose as polynomial criteria. The results are presented in terms of Antero Posterior, Proximo Distal, Medio Lateral Forces and Ankle Eversion, Plantar Flexion, Axial moments, discussing their role on the synovial lubrication phenomena effect in the Total Ankle Arthroplasty (TAR) for the optimal prostheses structural and tribological design.

1. Introduction
In the last year’s total joint arthroplasty (TJA) has become a common and a well-established surgical procedure in the case of severe arthritis especially regarding lower limb synovial joints [1]. Total Hip Replacements (THR) [2], Total Knee Replacement (TKR) [3] and Total Ankle Replacement (TAR) [4] are devoted to a substitution respectively of the hip, knee and ankle joint by using prostheses which unfortunately requires, in some cases, revisions and/or substitutions [5-8]. This problem is particularly felt in the case of THA since higher revision rate (about three times more for 100 patients) [9].

Among many others implants failure causes, nowadays particular interest is devoted to a correct tribological design of the implants [10, 11] in order to achieve more performant prostheses and to decrease, in this way, the rate of THA revision.

An optimized prostheses tribological design requires the choice of more and more performants materials in terms of stress, strain but also in wear resistance characteristics [12]; also the optimized
geometrical design is necessary to favour the synovial lubrication phenomena especially in terms of full film and/or elasto-hydrodynamic lubrication inside the bio-bearing gap [13,14].

For achieving this, a detailed description of the acting loads on the prostheses and of the kinematical behaviour during the common human daily activities is required. [15]. This is not always simple to obtain in-vivo, since direct force/displacements measurements on joints is in most cases not feasible due to economical and ethical problems [16] and it could be referred only to the body characteristics of the patients under experimental activity. For this purpose the possibility to obtain predicted loads from in-silico simulation is a challenge [17] toward an optimized wear assessment tool necessary for the optimized tribological design of the joint.

This paper aims to give a contribute toward this direction focusing on the theoretical background in which the musculoskeletal multibody algorithms operate, and discussing the obtained results with the use of a musculoskeletal multibody model with a Vaughan gait type.

2. Methods

2.1. The musculoskeletal model

In this study, we used a musculoskeletal modelling software, AnyBody Modelling System™ AMS, to estimate force and moment components acting on the ankle joint during a level walking. The AnyBody Modelling System™ is software based on the human musculoskeletal modelling able to simulate the dynamics of human motion. This environment adopts the inverse dynamics approach and different algorithms allow selecting the appropriate recruitment strategies allowing a complete analysis of the load components acting on the different joints of the human body during a known human body movement. Following the inverse dynamic approach, for achieving the joint forces the kinematic data and ground reaction force must be furnish as input for the simulations. These data are usually obtained in the Gait Analysis laboratories by using special Motion Capture apparatus, which allow to measure by cameras the subject’s gait kinematics, monitoring markers fixed in particular points of the body on the person’s skin (Figure 1).

![Figure 1. Four frames of gait model driven by kinematic data (The AnyBody System™)](image)

In particular this setup is able to measure an individual gait pattern by collecting the kinematic data of the lower limbs and the pelvis through a walking cycle (gait). Figure 1 shows three frames of the gait model driven by kinematic data. The dark spheres are the skin markers, the black lines representing ground reaction forces. The positions with their first and second derivatives in time, together with knowledge of the ground reaction forces, after a data filtering, represent the software input to predict the net forces in the leg. In fact, the inverse dynamics is based on the knowledge of the motion and the external loads data to determine the unknown internal forces. Following this approach, then the calculation of the behaviour of each muscle force is made possible by solving a redundancy muscular problem. In fact the muscular system is a quite complex system and for each motion many different sets of muscle forces could be involved; chose of the appropriate set is made by the central
nervous system (CNS) which instantly chooses one of them in order to produce the assigned kinematics. At moment the selection strategy is still not fully understand, however, the approach used by AnyBody Modelling System is well described in [18]; the software uses an algorithm to determine the activation of each muscle in order to replicate the function of the central nervous system. The list of the considered muscles in the model is reported in table 1.

| Muscle                        | Group     | Attachment |
|-------------------------------|-----------|------------|
| Soleus                        | Quadriceps| Shank      | Foot     |
| Gastrocnemius                 | Quadriceps| Thigh      | Foot     |
| Flexor Digitorum Longus       | Quadriceps| Shank      | Foot     |
| Flexor Hallucis Longus        | Quadriceps| Shank      | Foot     |
| Tibialis Posterior            | Quadriceps| Shank      | Foot     |
| Peroneus Brevis               | Quadriceps| Shank      | Foot     |
| Tibialis Anterior             | Quadriceps| Shank      | Foot     |
| Extensor Digitorum Longus     | Quadriceps| Shank      | Foot     |
| Extensor Hallucis Longus      | Quadriceps| Shank      | Foot     |
| Vastus Lateralis              | Quadriceps| Thigh      | Shank    |
| Vastus Medialis               | Quadriceps| Thigh      | Shank    |
| Vastus Intermedius            | Quadriceps| Thigh      | Shank    |
| Semitendinosus                | Hamstrings| Pelvis    | Shank    |
| Semimembranosus               | Hamstrings| Pelvis    | Shank    |
| Rectus Femoris                | Hamstrings| Pelvis    | Shank    |
| Biceps Femoris Caput Breve    | Hamstrings| Thigh    | Shank    |
| Sartorius                     | Hamstrings| Pelvis    | Shank    |
| Gracilis                      | Hamstrings| Pelvis    | Shank    |
| Iliopsoas                     | Hip muscles| Pelvis    | Thigh    |
| Gluteus Minimus 1             | Hip muscles| Pelvis    | Thigh    |
| Gluteus Minimus 2             | Hip muscles| Pelvis    | Thigh    |
| Gluteus Minimus 3             | Hip muscles| Pelvis    | Thigh    |
| Gluteus Medius 1              | Hip muscles| Pelvis    | Thigh    |
| Gluteus Medius 2              | Hip muscles| Pelvis    | Thigh    |
| Gluteus Medius 3              | Hip muscles| Pelvis    | Thigh    |
| Gluteus Maximus 1             | Hip muscles| Pelvis    | Shank    |
| Gluteus Maximus 2             | Hip muscles| Pelvis    | Shank    |
| Gluteus Maximus 3             | Hip muscles| Pelvis    | Thigh    |
| Tensor Fasciae Latae          | Hip muscles| Pelvis    | Shank    |
| Piriformis                    | Hip muscles| Pelvis    | Thigh    |
| Adductor Longus               | Hip muscles| Pelvis    | Thigh    |
| Adductor Magnus 1             | Hip muscles| Pelvis    | Thigh    |
| Adductor Magnus 2             | Hip muscles| Pelvis    | Thigh    |
| Adductor Magnus 3             | Hip muscles| Pelvis    | Thigh    |
| Quadratus Femoris             | Hip muscles| Pelvis    | Thigh    |
| Abductor Brevis               | Hip muscles| Pelvis    | Thigh    |
| Obturatorius Internus         | Hip muscles| Pelvis    | Thigh    |
| Obturatorius Externus         | Hip muscles| Pelvis    | Thigh    |
| Pictineus                     | Hip muscles| Pelvis    | Thigh    |
| Gemmuelus Inferior            | Hip muscles| Pelvis    | Thigh    |
| Gemmuelus Superior            | Hip muscles| Pelvis    | Thigh    |

Table 1. List of muscles included in the lower limb model
The used approach for solving the inverse dynamic problem accounting for the muscle recruitment strategy is based on an optimization problem.

Defining an objective function in the form:

\[
\min G(f^{(M)})
\]

In which \(f^{(M)}\) are the muscular forces for which

\[
0 \leq f_i^{(M)} \leq N_i, i \in \{1, \ldots, n^{(M)}\}
\]

The (2) states the non-negativity constraints on the muscle forces and that muscle can only pull (not push). The upper limit of the i-muscle strength capability is then assumed to \(N_i\).

Once defined the vector of the muscle forces and joint reactions in the form:

\[
f = [f^{(M)T} f^{(R)T}]^T
\]

The dynamic equilibrium equations can be obtained in the form:

\[
Cf = d
\]

where \(C\) is a coefficient matrix for the unknown forces/moments, while \(d\) is a vector of the known applied loads and inertia actions.

The most adopted objective function \(G\) forms, normalised for each muscle, are the polynomial criteria and the soft saturation criteria [19]:

\[
G(f^{(M)}) = \sum_{i=1}^{n^{(M)}} \left( \frac{f_i^{(M)}}{N_i} \right)^p
\]

\[
G(f^{(M)}) = -\sum_{i=1}^{n^{(M)}} \sqrt{1 - \left( \frac{f_i^{(M)}}{N_i} \right)^p}
\]

Both (5) and (6) contain a power variable \(p\) and a normalizing function for each muscle. In this study we used the approach (4) in which was settled \(p = 2\).

Ankle joint forces were simulated by using a 18 degrees of freedom lower limb made of 7 rigid members, the pelvis and the thigh, the shank and the foot (for each leg). From a kinematical point of view, the hip joint was assumed in the form of a spherical joint while the knee as a revolute joint and the ankle trochlear joint. In this study we used a set of kinematical input data from Vaughan et al. [20]. Human main parameters here adopted were a weight of 64.9 Kg and a height of 1.75.

3. Results

The output of the model will be presented in the in terms of load components acting on the ankle joint during the gait. With reference to the figure 2, the calculated loads are: the anterior-posterior force (Fx), the proximo-distal force (Fy), the medial-lateral force (Fz), the ankle eversion moment (Mx), the axial moment (My) and the plantar flexion (Mz).

![Figure 2. Reference system for ankle joint](image-url)
In figures 3, 4, 5 and 6 will be reported the results of the simulations in terms of forces and moment components.

Figure 3. Antero Posterior and Medio Lateral ankle joint components during the gait

Figure 4. Proximo distal ankle joint force component during the gait.

Figure 5. Ankle eversion and axial ankle moment components during the gait

Figure 6. Plantar flexion ankle moment component during the gait

About the muscle recruitment, the simulations allowed to calculate all the forces exerted by all the muscles considered in the lower limb model (Table 2) during the gait. Figure 4 shows a schematic image of the model which highlights the predominance during the toe-off the four muscles just listed. In figure 8 is reported a complete activation during the gait of the muscles involved. From an analysis of the obtained results is possible to observe that, despite the muscles involved during the movement of walking are numerous, more than 60% of the total force exerted during the 50% of gait cycle (toe-off) is provided by four major muscles, which are the Soleus (in the back part of the lower leg), the Gastrocnemius (in the back part of the lower leg), Rectus Femoris (one of the four quadriceps muscles of the human body), and the Iliopsoas (combination of the psoas major and the iliacus at their inferior ends). These muscles are distinct in the abdomen, but usually indistinguishable in the thigh.

Figure 7. Main active muscles during the gait
4. Discussion

It is well known that loading of the lower limb joints primarily depends on the physical activity (kinematical data) but they are also influenced by body weight (BW) but, in general, they individually differs greatly, even between subjects with the same BW. The simulations show maximum values of the ankle force and moment components in correspondence of about the 50% of gait cycle (toe-off phase) with a prevalence of the proximo distal force $F_y$ with a value in modulus of 2750 N and of the plantar flexion moment $M_z$ with a value of 82 Nm. The obtained behaviour of the loads, however, show good agreement with the others found in literature for example in [21]. The existing discrepancies however should be attributable to the fact that the considered model assume limited degree of freedom for the joint with the foot considered a single segment. Decreasing the degrees of freedom in the model allow a reduction in the computational time by reducing the complexity of the calculations to predict the muscle and joint contact forces, but on the other hand, this causes approximations in the force calculation which could accumulate on the whole kinematical chain furnishing more consistent discrepancies in the simulation.

Of course, another cause of alterations of the calculated forces is introduced by the anthropometric differences between the human bodies even if the scaling procedure aims to reduce it. Regarding the muscles activation Figure 8 shows the force exerted by each of the 42 muscles implemented in the model. As can be observed despite the muscles involved during the movement of walking are numerous, more than 60% of the total force exerted during the toe-off (50% of gait cycle) is provided by the four major muscles.

From a TKA design point of view, the obtained results allow the optimized design of the prostheses both from a structural point of view both from a tribological one [22, 23]. In fact the detailed knowledge of the load acting on the joint permits the accurate finite element modelling of the joint [24] to analyse the stability of the implant contributing to improve its stability and structural performances. Moreover the knowledge of the variation of the loads and of the kinematical quantities during the gait is necessary to the geometrical design of the synovial lubricated gap in order to achieve, according to Medley et al. [25] particular lubrication mechanisms (mixed or full-film) [22, 23]. This could be achieved by reaching an optimal value $h_{\text{min}}$ (minimum synovial meatus height).
divided by the root mean square of roughness values of the prostheses contact surfaces in order to optimize the prostheses performances in terms of wear resistance.

5. Conclusion
With the purpose of the optimization of the TKA design, a thorough determination of the loading of the ankle joint is necessary both for its stability and structural resistance design and for its tribological performances improvement, in terms especially of wear resistance. Unfortunately, in-vivo measurements of joint internal forces are not a simple and allowed task, and then non-invasive in-silico approach should furnish a meaning full perspective.

In this paper are presented ankle joint dynamical loading components during the gait, obtained by using the AnyBody Modelling SystemTM (AMS) computer software, adopting kinematical data obtained by Vaughan et al. [20] The obtained results, in terms of Antero Posterior, Proximo Distal, Medio Lateral Forces and Ankle Eversion, Plantar Flexion, Axial moments, shows a satisfying agreement allowing to be used both for detailed prostheses FEM analysis both for the optimized tribological design in terms of synovial lubricating mechanisms. Of course this investigation has limitations regarding the necessary full validation of the proposed model to be executed running several simulations, by varying key parameters of the model and by comparing the results with the ones (few) found in literature from in-vivo testing.

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