Knee Joint Energy Dissipation: A New Conceptual Approach To Inverse Dynamic Analysis

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***Abstract*— The purpose of this study was to investigate the effect of increasing the amplitude of contact forces at the knee on joint forces and moments at the hip level. This was done to account for dissipative effects occurring at the knee articular cartilage. Knee joint reaction forces and moments were computed using 3D inverse dynamics for two healthy subjects and one grade IV osteoarthritic subject during a steady-state walking on treadmill at selected self-pace. A 2D contact model of the knee was used in order to obtain the tibio-femoral contact forces from the net joint forces in the sagittal plane. To account for energy dissipation, dissipative force vectors were added to tibio-femoral contact forces and transferred back into joint distal forces of the adjacent segment. The latter were inputted recursively into inverse dynamics program to compute the net joint forces and moments at the hip. Significant differences were observed in the peaks for proximal/distal joint force and medial/lateral joint moment at the hip. The results obtained in this study are preliminary and consistent with our hypothesis and suggest that energy dissipation could have a significant impact on the results of an inverse dynamic analysis.**

Keywords-component; inverse dynamics, energy dissipation, contact forces, knee articular cartilage.

I. Introduction

The inverse dynamic approach has been widely used in the literature to compute joint forces and moments in the lower limbs for different daily activities such as gait and arising from a chair [1-3]. It requires segment kinematics, body segment parameters and ground reactions as input data. These data are fitted into Newton-Euler equations of motion to obtain net joint forces and moments. Models of the lower limbs using the inverse dynamic approach are limited in several aspects and their validity depends on simplifying assumptions.

One of the fundamental assumptions used in inverse dynamics is that joint energy dissipation is assumed to be negligible. By assuming this, the joint reaction forces and moments estimated at the proximal end of the segment are totally transmitted in amplitude and direction to the distal end of the adjacent segment as an action. Nevertheless, it has been proven in the literature that, due to its multiphasic nature, articular cartilage dissipates non-negligible quantities of energy when subjected to compressive loads [4-5]. According to the biphasic theory of Mow et al. [4], articular cartilage has a solid phase and a fluid phase. The solid phase is mainly composed of collagen fibers and proteoglycans which form a porous and permeable solid matrix. On the other hand, the fluid phase is composed of an interstitial fluid which lies within the solid matrix. Together, the solid phase and the fluid phase are said to be intrinsically incompressible since deformation of the solid matrix is possible only if there is fluid exudation out of the cartilage. When this fluid exudation occurs, the relative velocity of the interstitial fluid with respect to the velocity of the solid matrix causes frictional drag and hence energy dissipation occurs [4-6]. This dissipative effect depends on the permeability of articular cartilage and is not negligible for healthy articular cartilage [7]. However, it has been shown that fractional drag is negligible for osteoarthritic articular cartilage [7].

The viscoelastic behaviour of the solid matrix due to the interaction between its components also contributes to dissipate energy when articular cartilage is compressed or sheared. Nevertheless, for healthy articular cartilage subjected to a compression, it has been shown that this dissipation is negligible compared to the dissipation due to frictional drag but becomes more important in the case of osteoarthritic cartilage [7]. Also, for joint rotations, the sliding of two articular surfaces leads to energy dissipation. This dissipation is proportional to a time-dependent friction coefficient and tends to increase in the case of osteoarthritic knee articular cartilage [8-10].

These dissipative mechanisms are important since they characterise the mechanical behaviour of articular cartilage and they should not be neglected. Their effect can be integrated in an inverse dynamic analysis if the contact forces between the segments are known. Few models of the joints of the lower limbs that compute contact forces from joint forces obtained from inverse dynamic computations were found in the literature [11-13]. Contact forces obtained from these models have usually a single point of application and this point is not necessarily located on the contact region defined as the physical contact area between two adjacent segments. In 1968, Morrison [11] was the first to develop a 3D contact model of the knee joint. However, the model of Morrison requires the knowledge of several physiological parameters difficult to obtain. More recently, Li [12] developed a 2D model of the knee that uses knee kinematics, muscle forces and body segment parameters to obtain tibio-femoral contact forces in the sagittal plane. This model is relatively simple and the inverse procedure is possible: i.e. knee joint forces could be estimated from knee contact forces only if the knee joint moments are known.

It is hypothesised here that energy dissipation estimated from the different knee articular cartilage dissipative mechanisms can be combined into three orthogonal dissipative force vectors that have the same orientation and origin than the knee contact forces. Therefore, these vectors can be summed and transformed back into joint forces using a contact model of the knee. The knee joint forces obtained can be integrated recursively in an inverse dynamic analysis to enhance the hip joint forces and moments. To date, no in vivo experiments have been made to measure energy dissipation related to knee articular cartilage for both static and dynamic conditions. The complexity of its geometry and its biological composition makes difficult the estimation of dissipative effects. However, one thing that can be estimated is the direction of the dissipative force vectors. Considering the case of the knee during gait, the femur acts like a piston compressing and shearing articular cartilage over the tibial surface. The force exerted by the femur on articular cartilage is prescribed by the weigth of the trunk, of the upper limbs and of the head and also by their inertia. However, the resulting force transferred to the tibia is expected to be less than the resulting force applied by the femur on articular cartilage when energy dissipation occurs. Now, considering an inverse dynamic analysis, the contact forces in this case are transferred from the tibia to the femur. Therefore, to account for energy dissipation, the dissipative forces should be added to the contact forces.

The purpose of this study is to analyse the effect of increasing contact forces at the knee on the joint forces and moments at the hip which are calculated using the inverse dynamic approach. Different levels of energy dissipation will be tested. Furthermore, the differences in force and moment patterns at the hip will be investigated for both healthy and osteoarthritic knee articular cartilage.

II. METHODS

A. Inverse Dynamic Method

Knee joint forces and moments were obtained from a generic inverse dynamic method developed by Dumas et al. [14]. This method uses quaternion algebra instead of Euler/Cardanic angles to overcome kinematic formalism. Quaternion algebra has the advantages of not being sequence-dependent and also of not suffering from singularities caused by gimbal lock [14]. In addition, the generic method of Dumas uses wrench notation which allow the simultaneous computations of forces and moments expressed at a predefined point location on a segment. The knee joint forces and moments computed are expressed in the inertial coordinate system (ICS).

B. Experimental setup

Inverse dynamic output data from two healthy subjects, one of 24 years old (subject 1) and one of 80 years old (subject 2), and from one grade IV osteoarthritic subject (60 years old; subject 3) have been used in this study. Briefly, all subjects were asked to walk at a selected self-pace on a tread mill equipped with a force-platform. Four markers fixed on rigid body were placed on each segment (thigh, shank and foot) by mean of exoskeletal system [15] and the data acquisition was performed using a Vicon M460 system (Vicon Peak Inc.) equipped with six cameras at 120 Hz. The accuracy of this system is 0.5 mm. The data acquisition of the force plate was synchronised with the data acquisition of the markers. For each trial, 17 to 20 gait cycles per subject were recorded.

C. Computation of Contact Forces

The joint forces obtained from inverse dynamic method were then transformed into tibio-femoral contact forces using the procedure used by Costigan [16] which is based on the 2D contact model developed by Li [12] in the sagittal plane. Prior to transformation, the joint forces which were expressed in the ICS were computed in the segment coordinate system (SCS) defined at the proximal end of each segment. The muscle groups involved in this model are the quadriceps and the hamstrings. The contribution of the gastrocnemius was not considered since the acquisition of its electromyographic measure is not repeatable due to calibration problem. Also, the force generated by the quadriceps was assumed to be totally transferred into the patellar ligament.

The muscle forces were calculated from the net joint moments at the knee by using their corresponding lever arms and by assuming no antagonistic activity. The lever arms, which correspond to the smallest distance between the knee joint center and the line of action of the muscle groups, were derived from the knee joint angle by using regression equations as proposed by Smidt [17]. It was also assumed that the position of the knee joint center remains fixed during flexion/extension movements. During knee extension, the moment exerted by the patellar ligament on the tibio-femoral joint was set to zero and, during knee flexion, the moment exerted by the hamstrings on the tibio-femoral joint was also set to zero. By applying this, the set of equations of motion in the sagittal became determinate and the muscle forces were obtained straightforward.

Anterior/posterior (AP) and proximal/distal (PD) contact forces were calculated using Newton's second law of motion. The medial/lateral (ML) contact forces were assumed to be identical to the ML joint forces.

D. Accounting for Energy Dissipation

To account for energy dissipation, dissipative forces were first defined. They were expressed as a percentage (α) of the contact forces in the AP, PD and ML directions:

$$FD_i = \alpha \cdot F_i; \qquad i = 1,2,3 \tag{1}$$

where FD and F are the dissipative and contact forces respectively in the i^{th} direction. Different values of (α) were tested: 0%, 5%, 10%, 15% and 20%. The dissipative forces obtained from (1) were added to the contact forces to form the new modified contact forces. Afterwards, the modified joint forces were calculated using the contact model described previously and by applying the procedure inversely. These joint forces, which were expressed in the SCS, were transferred in the ICS using quaternion algebra and were inputted recursively into the generic inverse dynamic method. Joint forces and moments obtained at the hip were resampled according to a generic gait cycle and were then averaged. For all subjects and levels of energy dissipation, the hip joint forces and moments were computed and analysed.

III. Results

To analyse the effect of energy dissipation occurring at the knee joint on the net hip joint forces and moments, the percentages corresponding to the increases in hip joint forces and moments at the instant of peak were calculated relative to a baseline that correspond to 0% level of energy dissipation. Table I depicts the percentages of the PD hip joint force and the ML hip joint moment for all subjects. In table I, it is shown that a 10% increase in energy dissipation corresponds approximately to a 20% increase in the PD hip joint force for the three subjects. Similarly, a 20 % increase in energy dissipation results in a 40% increase in PD knee joint force. For the ML hip joint moment, large discrepancies among the subjects have been observed for all increases in energy dissipation. Subject 1 obtained the largest increases in ML hip joint moment as apposed to subject 3 who obtained the smallest values. Increases in hip joint forces and moments in other directions (not shown) were lower than those presented in Table I.

Fig. 1 shows the ML hip joint moment for all three subjects as a function of the gait cycle starting at heel strike. It is observed from the graphs of subjects 1 and 2 that larger discrepancies occur at the beginning of the gait cycle. Subject 3 exhibits a different pattern since the increase due to the dissipative energy remains constant throughout all the gait cycle. This is not surprising since subject 3 is an osteoarthritic subject with high degree of pathology.

TABLE I. Effects of energy dissipation on net hip joint forces and moments at the instant of peak

Subject	Levels of energy dissipation compared	Increase in PD force (%)		Increase in ML moment (%)	
		μ	***SD***	***μ***	***SD***
1	5%	9.6	0.3	16.4	2.0
	10%	19.2	0.5	35.7	4.4
	15%	28.8	0.8	53.4	5.1
	20%	38.4	1.1	74.7	6.3
2	5%	9.9	0.4	11.6	0.8
	10%	19.7	0.9	23.2	1.5
	15%	29.6	1.3	35.9	3.6
	20%	39.4	1.9	49.1	4.3
3	5%	9.0	0.8	5.7	1.8
	10%	18.0	1.7	11.4	3.2
	15%	27.0	2.6	17.1	5.6
	20%	36.0	3.5	22.9	7.6

μ and SD correspond to the mean and the standard deviation respectively

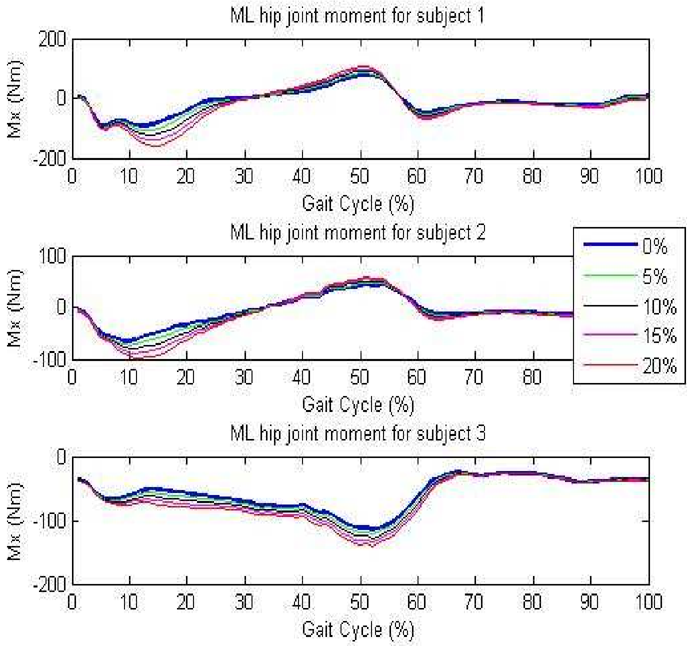

Figure 1. ML hip joint moment for all subjects and for different levels of energy dissipation incorporated to knee contact forces

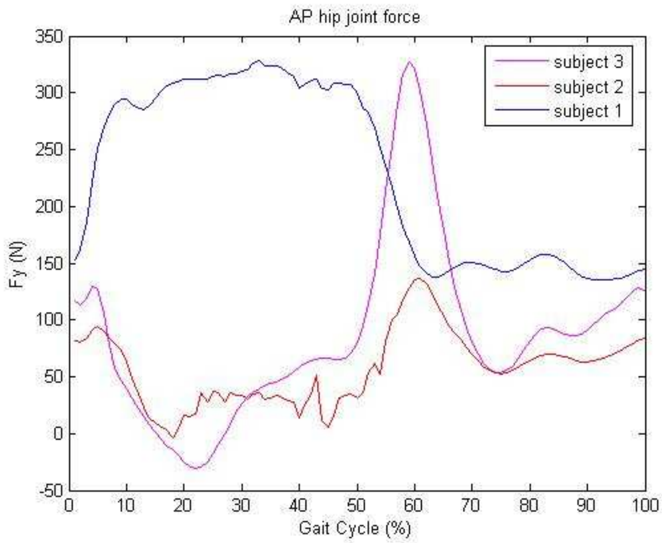

Figure 2. AP hip joint force for all subjects and for a level of energy dissipation of 0%

IV. Discussion

The purpose of this study was to determine whether accounting for energy dissipation has a significant effect on the force and moment estimates at the adjacent segment. The results obtained in this study show that this effect is not negligible for constant energy dissipation levels greater than 5%. There exists a linear relation between the increase in hip joint forces and the levels of energy dissipation. Doubling the amount of energy dissipated at the knee results in doubling the increase in hip joint forces. This is due to the linearity of the balance of forces in Newton-Euler equations. On the other hand, the relation between the increase in hip joint moments and the levels of energy dissipation compared is nonlinear. This nonlinearity depends on the value of the lever arm used in inverse dynamics which is defined as the vector relating the knee articular joint center to the hip articular joint center. The standard deviations associated with the increases in hip joint forces and moments depend on the repeatability of the peak values of the hip joint forces and moments. It has been noticed that the standard deviations tend to increase as the energy dissipation level increases. This is true since the introduction of a larger noise leads to a greater dispersion in the data.

From fig. 1, it seems that higher amplitude in hip joint peak moments for the 0% level of energy dissipation lead to an increase of the gap between the hip joint peak moments for the other levels of energy dissipation. In fact, the larger are the knee joint forces the larger the gap between the different curves representing energy dissipation will be. For subject 3, this explains why the discrepancies between the curves remain approximately constant during the stance phase since the amplitude of the AP knee joint force (fig. 2) is relatively important compared to the other subjects and remains constant in amplitude.

The methodology used in this study is fairly simple but is limited in several aspects. First of all, the knee contact model used in this study is 2D and hence does not compute the ML knee contact force. To overcome this problem, the ML knee contact force was assumed to be equal to the ML knee joint force. This assumption should have had little effects on the results obtained since its amplitude range is much smaller than the amplitude range of the PD knee contact force.

The knee contact forces were calculated at the knee joint center instead of calculating them at their point of application on the tibial surface. By doing this, there is no moment associated with the creation of the dissipative force vectors. This assumption could lead to imprecision in the computation of hip joint moments since knee joint moments should ideally account for the change in knee contact forces. The number of subjects is also an important limitation in this study. A larger number of subjects in each cohort could have highlighted some characteristic trends in the results obtained notably for the osteoarthritic cohort.

The dissipative effects were assumed to be constant in this study. In fact, it is well known that the energy dissipation at the knee is more likely to vary during the gait cycle [18]. This energy dissipation is dependent on the magnitude and the direction of the forces applied on the knee articular cartilage and is dependent also on several physical and mechanical parameters [4]. Realistic estimates of the energy dissipation occurring at the knee require the use of a model of the knee articular cartilage that would include all the parameters governing its mechanical behaviour. The development of such a model could improve significantly the results obtained in this study. Although, this study is limited in several aspects, it has shown that non-negligible levels of energy dissipation at a joint (i.e. greater that 5%) have a considerable impact on the forces and moments obtained from inverse dynamics at the proximal end of the adjacent segment.

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