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Loading applied on prosthetic knee of transfemoral amputee: comparison of inverse dynamics and direct measurements

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Abstract

Inverse dynamics is the most comprehensive method that gives access to the net joint forces and moments during walking. However it is based on assumptions (i.e., rigid segments linked by ideal joints) and it is known to be sensitive to the input data (e.g., kinematic derivatives, positions of joint centres and centre of pressure, inertial parameters). Alternatively, transducers can be used to measure directly the load applied on the residuum of transfemoral amputees. So, the purpose of this study was to compare the forces and moments applied on a prosthetic knee measured directly with the ones calculated by three inverse dynamics computations - corresponding to 3 and 2 segments, and « ground reaction vector technique » - during the gait of one patient. The maximum RMSEs between the estimated and directly measured forces (i.e., 56 N) and moment (i.e., 5 N.m) were relatively small. However the dynamic outcomes of the prosthetic components (i.e., absorption of the foot, friction and limit stop of the knee) were only partially assessed with inverse dynamic methods.

Keywords: inverse dynamics, ground reaction vector technique, direct measurements, transfemoral amputation, prosthetic gait

1. Introduction

The evaluation of the dynamic loading on the prosthetic limb is critical during rehabilitation program (e.g., load bearing exercises, effect of walking aids) and fitting of prosthesis (e.g., alignment, choice of component, socket discomfort [1-3]).

Inverse dynamics is the most comprehensive method that gives access to the net joint forces and moments associated with the load bearing on prosthetic components (i.e., knee, foot, protective device). This method is based on assumptions that rigid segments are linked by ideal joints. Besides, it is known to be sensitive to input data such as inertial parameters [4], kinematic derivatives, positions of joint centres and centre of pressure. Results are only partially representative because of limited number of steps, level walking, forceplate targeting, etc, unless costly instrumented treadmill are used.

Alternatively, portable kinetic systems relying on transducer have been developed to measure directly the load applied on the residuum of transfemoral amputees fitted with sockets [5] or osseointegrated fixation [6] during activities of daily living. So far, a side-by-side assessment of both methods is yet to be performed.

The purpose of this study was to compare the forces and moments applied on a prosthetic knee measured directly with the ones calculated by three inverse dynamics computations [7] corresponding to 3 and 2 segments, and « ground reaction vector technique » [8, 9].

2. Material and Methods

One fully rehabilitated female transfemoral amputee (36 yr, 1.6 m, 62.6 kg) participated in the study. The research institution's human ethics committee approved this study. The participant provided informed written consent.

She walked at a self selected speed with a prosthesis including a socket, a multiaxial transducer (JR3 Inc., Woodland, CA, USA), and her usual prosthetic knee, foot and footwear (Figure 1a). The three components of forces and moments were measured with accuracy better than 1 N and 1 N/m, respectively using the transducer and recorded by a laptop at 200Hz [5]. The position and orientation of the transducer were aligned with the socket. The gait data were recorded simultaneously with a 6-camera Peak-Motus (VICON, Oxford, UK) and a forceplate (AMTI, Watertown, MA, USA) at 50Hz and 500Hz, respectively. The transducer and gait data were manually synchronized a posteriori, using the Superior-Inferior force at heel contact. An error of ± 1 frame (at 200 Hz) introduced root mean square errors (RMSEs) of 2 N, 4 N, 13 N and of 2 N.m, 1 N.m, 1 N.m in the three components of forces (i.e., Lateral-Medial, Anterior-Posterior and Superior-Inferior) and moments (i.e., Extension-Flexion, Adduction-Abduction and Internal-External Rotation), respectively.

Segment Coordinate Systems (SCSs) were constructed from markers placed on the socket, transducer, pylon and shoe, approximately at landmark levels (i.e., great trochanter, tibial tuberosity, calcaneum, 5th metatarsal head) and on mechanical parts (i.e., knee axis, ankle fixation). The inertial parameters of the prosthetic limb were estimated using volume of residuum and bench top measurements of each component.

The forces and moments were computed by 3D inverse dynamics [7] at the ankle fixation, midpoint on the knee joint axis and hip joint centre. Both knee joint axis and hip joint centre were estimated by functional methods. Two segmentations of the limb corresponding to 3 and 2 segments (Figures 1b and 1c) were evaluated since the prosthesis did not include an articulated ankle but a deformable heel providing a pseudo-plantar flexion after heel strike. The « ground reaction vector technique » [8] was also evaluated giving that the inertial parameters of the prosthetic segments might be considered negligible, as suggested in some studies [9]. No free body diagram is considered as for static or dynamic computations (Figure 1d): the ground reaction force and moment are simply transformed from the forceplate to joint centres using classical rigid body mechanics. Therefore, the 3D force remained the same while the force times the lever arm from its line of action was added to the 3D moment. The opposite sign was considered in order to represent the internal action of the proximal adjacent segment as it was defined in inverse dynamics computations [7].

The forces and moments obtained with inverse dynamics, ground reaction vector technique and direct measurements were compared for the knee, expressed in the thigh SCS, using RMSEs. To do so, the force and moment measured directly with the transducer were transformed to knee joint centre as previously done for the ground reaction force and moment. It was assumed that the inertial parameters of the short connector between the transducer and the prosthetic knee were negligible. The opposite sign was also considered.

3. Results

The participant walked at 1.21 m/s. The support and swing phases represented 57% and 43% of the gait cycle, respectively.

The knee joint forces and moments, and the RMSEs between the estimated (i.e., inverse dynamics and ground reaction vector technique) and directly measured forces and moments are presented in Figure 2 and Table 1, respectively. Only slight differences were found between the inverse dynamic computations considering either 3 or 2 segments. Conversely, the ground reaction vector technique provided the most divergent curves with the highest RMSE on Superior-Inferior force during support and systematically null force and moment during swing.

Nevertheless, the dynamic outcomes of the prosthetic components were partially assessed, regardless of the computation method. On one hand, the outcomes of the deformable foot could be observed in the Superior-Inferior force at initial stance. Conversely to the ground reaction vector technique, the prosthetic segment weights were rigorously subtracted to the ground reaction force in the inverse dynamic computations. However, the direct measurement revealed a smaller force corresponding to an absorption in the foot. On the other hand, the outcomes of the mechanism could be observed in the Superior-Inferior force and Extension-Flexion moment at initial and terminal swing. While the ground reaction vector technique assumed null force and moment, the derivatives of linear and angular momentums were evaluated in the inverse dynamic computations. However, the direct measurement revealed vibrations and impacts corresponding to the friction and limit stop of the knee, respectively.

4. Discussion

The pattern and magnitude of the knee joint forces and moments estimated by the inverse dynamics and the ground reaction vector technique and directly measured were typical of transfemoral amputees [2, 4-6].

However, the ground reaction vector technique provided the most divergent curves. This technique had been strongly criticized in 2D [8] and it should be preferred even less in 3D. Conversely, the inverse dynamics with either 3 or 2 segments provided comparable curves. The question of modelling the prosthetic limb with either 3 or 2 segments has risen because considering 3 segments might be a way to cope with a deformable foot. Moreover, whatever the prosthesis is, an ankle joint is almost always considered in the literature [1, 3, 4] in order to match up to the sound limb and to compare with the asymptomatic gait. Even if this modelling choice had little influence on the knee results, the significance of displaying and interpreting joint moment, and to a greater extent joint power [3], at an ankle level with no existing joint remains questionable.

Overall, the maximum RMSEs between the estimated and directly measured forces (i.e., 56 N) and moment (i.e., 5 N.m) were reasonably small. This might be due to the lack of soft tissue artefacts except, to some extent, for the prosthetic foot and the shoe. However, the application of inverse dynamics to prosthetic gait demonstrated typical errors dealing with the assumptions of rigid segments and ideal joints and with the sensitivity kinematic derivatives. The prosthetic components are designed to provide absorption at the foot and friction and limit stop at the knee. These dynamic outcomes were only partially assessed: the Superior-Inferior force was over-estimated at initial stance, and both Superior-Inferior force and the Extension-Flexion moment were under-estimated during swing.

One limitation of the study is the assessment of only one transfemoral amputee. The comparison of the knee joint forces and moments estimated by the inverse dynamics and directly measured could be extended to other prosthetic designs, providing other dynamics outcomes (e.g. pneumatic and hydraulic prosthetic knee [2]). Another limitation is the kinematic frequency acquired at only 50 Hz. Other inverse dynamic computations based on accelerometers [10] designed to cope with high-speed dynamics, might provide better insight. However, these methods have yet to be applied to prosthetic gait. Moreover, the synchronisation with more classical motion analysis systems is not always straightforward as it was the case in our study for the transducer.

Future inverse dynamic computations based on direct measurements other than the ground reactions will allow to better estimate the hip joint forces and moments on prosthetic side taking into account the dynamics of the absorption, friction and limit stop of the prosthesis components.

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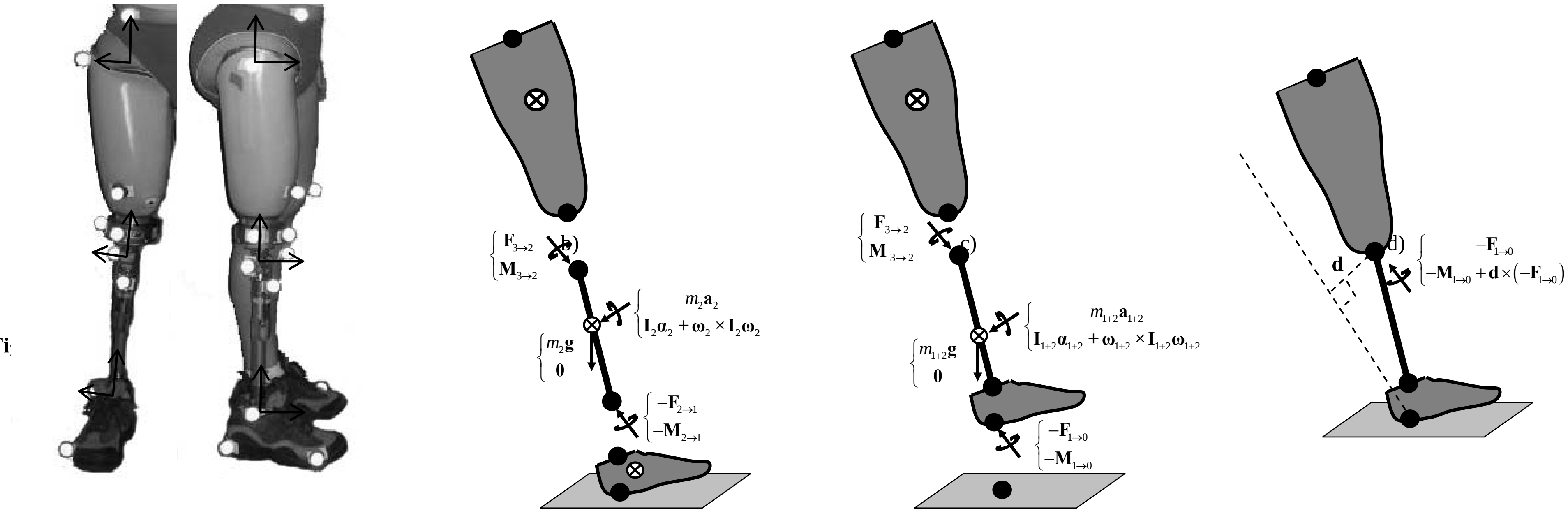
Table and figure captions

Figure 1:
a) Prosthetic limb with multiaxial transducer and motion analysis markers allowing constructing the segment coordinate systems
b) Free body diagram (in wrench notation [7]) for inverse dynamics considering 3 segments. Indices i = 1, 2 and 3 stand for foot, leg and thigh. The joint force and moment of the proximal adjacent segment $\begin{cases} \mathbf{F}_{i+1\to i} \\ \mathbf{M}_{i+1\to i} \end{cases}$ are computed from the segment mass m_i, inertia matrix $\mathbf{I}_i$, linear acceleration of centre of mass $\mathbf{a}_i$, angular velocity and acceleration ω_i and α_i , the gravity acceleration $\mathbf{g}$ and the joint force and moment of distal adjacent segment.
c) Free body diagram (in wrench notation [7]) for inverse dynamics considering 2 segments. Indice i = 1+2 stands for a « foot and leg » segment and indice i = 0 stands for the forceplate.
d) Principle of « ground reaction vector technique » [8, 9]. The external force times the lever arm **d** from its line of action is added to the external moment to approximate the joint moment.

Figure 2:
Knee joint forces and moments estimated by 3D inverse dynamics (considering either 3 or 2 segments), estimated by ground reaction vector technique and directly measured by multiaxial transducer, expressed in the thigh SCS (Lateral-Medial, Anterior-Posterior, Superior-Inferior directions, and Extension-Flexion, Adduction-Abduction, Internal-External Rotation orientations).

Table 1:
Root Mean Squared Errors (RMSEs) between the estimated and directly measured knee joint forces and moments during support and swing phase.

Figure 1


F3→2
M3→2
b)
m2a2
I2α2 + ω2 × I2ω2
m2g
0
−F2→1
−M2→1
F3→2
M 3→2
c)
m1+2a1+2
I1+2α1+2 + ω1+2 × I1+2ω1+2
m1+2g
0
−F1→0
−M1→0
d)
d
−F1→0
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Fi

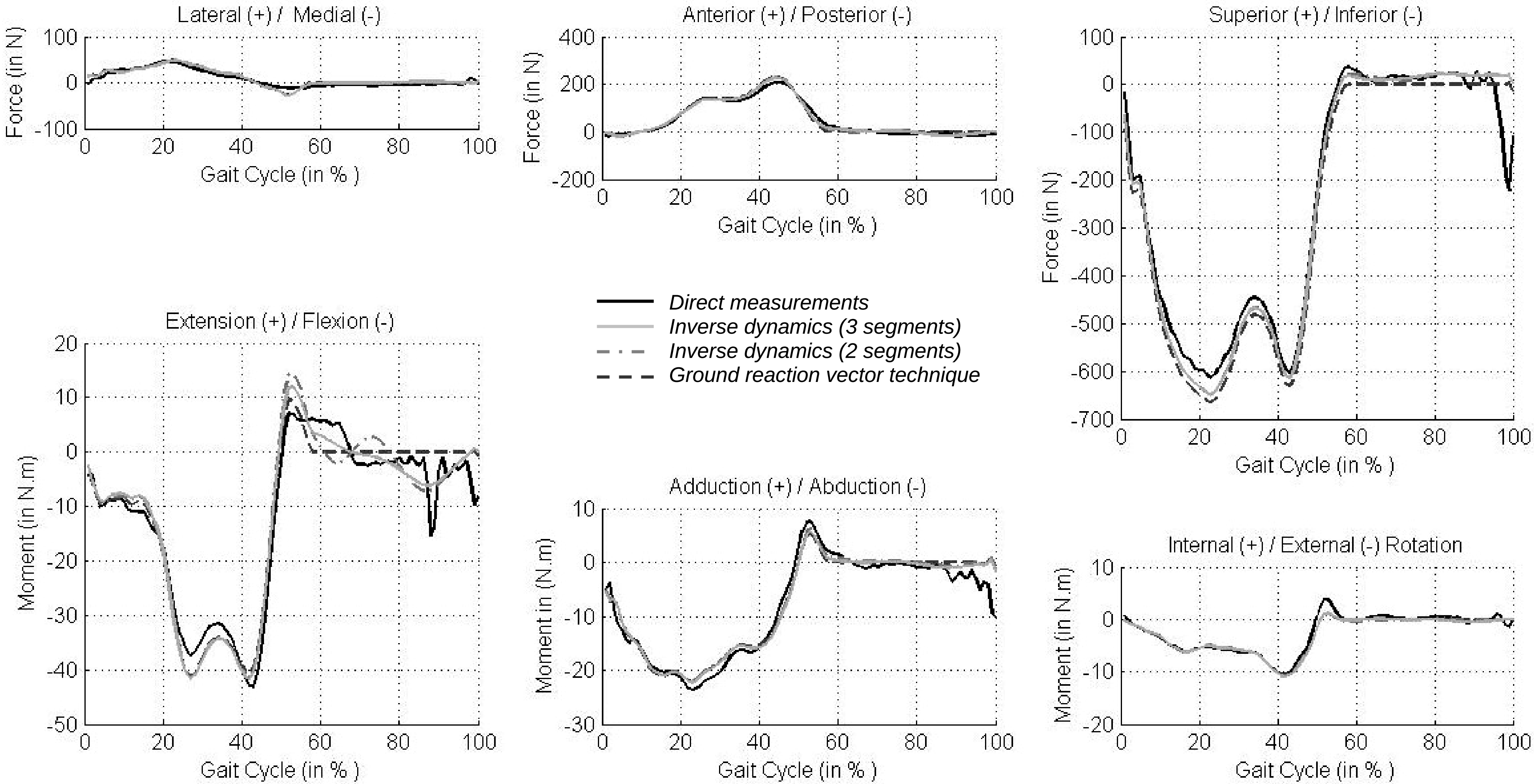
Lateral (+) / Medial (-)
Force (in N)
Gait Cycle (in %)
Anterior (+) / Posterior (-)
Force (in N)
Gait Cycle (in %)
Superior (+) / Inferior (-)
Force (in N)
Gait Cycle (in %)
Extension (+) / Flexion (-)
Moment (in N.m)
Gait Cycle (in %)
Direct measurements
Inverse dynamics (3 segments)
Inverse dynamics (2 segments)
Ground reaction vector technique
Adduction (+) / Abduction (-)
Moment in (N.m)
Gait Cycle (in %)
Internal (+) / External (-) Rotation
Moment (in N.m)
Gait Cycle (in %)

Table 1

			Inverse dynamics (3 segments)	Inverse dynamics (2 segments)	Ground reaction vector technique
Support	Force (in N)	Lateral-Medial	7	5	7
		Anterior-Posterior	11	11	15
		Superior-Inferior	26	26	42
	Moment (in N.m)	Extension-Flexion	3	3	3
		Adduction-Abduction	2	1	2
		Internal-External Rotation	1	1	1
Swing	Force (in N)	Lateral-Medial	7	6	5
		Anterior-Posterior	8	9	11
		Superior-Inferior	56	55	52
	Moment (in N.m)	Extension-Flexion	4	5	5
		Adduction-Abduction	2	3	3
		Internal-External Rotation	1	1	1