Joint Contact Forces can be Reduced by Improving Joint Moment Symmetry in Below-Knee Amputee Gait Simulations

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Abstract

Despite having a fully functional knee and hip in both legs, asymmetries in joint moments of the knee and hip are often seen in gait of persons with a unilateral transtibial amputation (TTA), possibly resulting in excessive joint loading. We hypothesize that persons with a TTA can walk with more symmetric joint moments at the cost of increased effort or abnormal kinematics. The hypothesis was tested using predictive simulations of gait. Open loop controls of one gait cycle were found by solving an optimization problem that minimizes a combination of walking effort and tracking error in joint angles, ground reaction force and gait cycle duration. A second objective was added to penalize joint moment asymmetry, creating a multi-objective optimization problem. A Pareto front was constructed by changing the weights of the objectives and three solutions were analyzed to study the effect of increasing joint moment symmetry. When the optimization placed more weight on moment symmetry, walking effort increased and kinematics became less normal, confirming the hypothesis. TTA gait improved with a moderate increase in joint moment symmetry. At a small cost of effort and abnormal kinematics, the peak hip extension moment in the intact leg was decreased significantly, and so was the joint contact force in the knee and hip. Additional symmetry required a significant increase in walking

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effort and the joint contact forces in both hips became significantly higher than in able-bodied gait.

Keywords: Biomechanics, Simulation, Human Locomotion, Below-knee Amputee, Optimal Control

1. Introduction

The prevalence of limb amputation is expected to rise to one in 95 Americans in 2050. Most amputations are transtibial amputations due to complications of diabetes. However, amputations due to trauma are more common in young and physically active individuals (1). Persons with a transtibial amputation (TTA) who have worn a prosthesis for over five years have a higher occurrence rate of osteoarthritis in the knee and hip of the intact leg (2; 3). Also, other health issues exist among persons with an amputation, such as residual limb pain, back pain and phantom pain (4).

Studies analyzing gait of persons with a TTA have reported asymmetries between the residual and intact leg in the joint moments, the stance and swing time, the ground reaction force (GRF) (5) and EMG data (6). These asymmetries, especially in the joint loading, may cause higher forces in the intact leg compared to the residual leg and to able-bodied gait, which is suspected to be related to the aforementioned higher occurrence rate of osteoarthritis (2). Therefore, it is desirable to reduce the level of asymmetry in the joint moments

17 of the knee and hip in TTA gait and distribute the load more equally between
18 both legs.

19 The asymmetries are presumably caused by compensations required after
20 amputation and the resulting loss of the biological ankle function (7; 8). Elastic
21 storage and return (ESAR) feet were developed to improve TTA gait (9; 10).
22 However, the joint moments remain distinctly asymmetric when using the ESAR
23 feet (11; 12; 8).

24 Asymmetry is inevitable because the prosthetic limb does not replicate the
25 mechanics and control of the intact limb. However, the knee and hip, and
26 the entire intact limb, still have a sufficient set of muscles under voluntary
27 neural control for flexion and extension. Therefore, we hypothesize that persons
28 with a TTA can walk with symmetric joint moments using a passive prosthesis,
29 but prefer not to do so because it would require additional effort or abnormal
30 kinematics.

31 If humans choose their gait to minimize energy (13), this question can be
32 investigated by predictive simulations of gait. In such simulations, an optimal
33 control problem is formulated to find open loop controls that produce walking
34 with minimal cost. Previous work showed that the main features of gait can
35 be predicted by minimizing muscle activation (14; 15), and that predictions
36 are improved when tracking of joint angles, GRFs and duration of normal gait
37 is added to the objective (16). This objective has been used to predict the
38 effect of sports equipment on performance (16) which is conceptually similar
39 to predicting TTA gait, because both cases predict how an altered mechanical
40 environment affects gait.

41 In the present study, a second objective is added to penalize asymmetry in
42 joint moments. The second objective creates a multi-objective optimization,
43 which is used to study the effect of different levels of joint moment symmetry
44 on TTA gait.

2. Methods

2.1. Model

A sagittal plane model of an able-bodied human (Figure 1) is used. It has seven rigid segments and nine degrees of freedom: the position and orientation of the trunk, and the hip, knee and ankle joint angles (16). Dimensions and inertial properties were taken from Winter (17). Eighteen state variables are associated with the multibody dynamics: nine generalized coordinates and nine generalized velocities.

Sixteen muscles actuate the model in the able-bodied configuration. These muscles were modeled as three-element Hill type muscles with a contractile element with activation dynamics, force-length and force-velocity relationships, and a parallel and series elastic element. The muscle properties were the same as in (16). The muscle fiber length and activation state of each muscle, 32 states, were added to the state of the system. When TTA gait was simulated, the gastrocnemius, soleus and tibialis anterior muscle were removed in one of the legs and replaced by a passive spring with a stiffness of 600 Nm/rad and damping of 15 Nm/s/rad, as used by Zmitrewicz et al. (18). The mass and inertial properties were not altered.

Contact between the foot and the ground was modeled using four contact points, one at the heel and one at the toe of each foot. The vertical GRF, F_y depends on the vertical position y of the contact point, as follows:

$$F_y(d) = kd(1 - by) - k_g y \quad (1)$$

The first term represents a visco-elastic force due to a deformation, d , of the contact point. The second term is a weak linear spring to aid the optimization. To ensure differentiability, d was calculated as

$$d = \frac{1}{2} \cdot \left(\sqrt{y^2 + y_0^2} - y \right) \quad (2)$$

where y_0 is the size of the transition region between contact and no contact. The following parameter values were used: $k = 2 \cdot 10^4$ N/m, $b = 1$ s/m, $k_g = 1$ N/m, $y_0 = 1 \cdot 10^{-3}$ m.

74 The horizontal GRF was modeled as a continuous approximation of Coulomb
75 friction:

$$76 \quad F_x(f_y, \dot{x}) = -\mu F_y \frac{\dot{x}}{\sqrt{\dot{x}^2 + v_0^2}} \quad (3)$$

77 where $\dot{x}$ is the sliding velocity of the contact point, $\mu = 1$ is the friction coefficient
78 and $v_0 = 0.1$ m/s is a small velocity parameter that ensures a sign change in
79 the friction when there is a change in the sliding direction.

80 Van den Bogert et al. (16) derived the dynamics equations for the multi-
81 body system with the ground contact model using Autolev (Online Dynamics,
82 Sunnyvale, CA) with respect to the nine generalized coordinates q :

$$83 \quad M(q) \cdot \ddot{q} + B(q, \dot{q}) - \tau = 0 \quad (4)$$

84 $M(q)$ is the mass matrix and $B(q, \dot{q})$ a vector that contains the Coriolis and
85 gravity forces, the GRFs, and passive joint torques. The first three elements of
86 the generalized force vector, τ , are zero since these are unactuated degrees of
87 freedom. The six joint moments, τ_4 to τ_9 , are related to the muscle forces via
88 the muscle moment arms, d_{ij} using equation 5.

$$89 \quad \begin{bmatrix} \tau_4 \\ \vdots \\ \tau_9 \end{bmatrix} = \begin{bmatrix} d_{1,1} & d_{1,2} & \dots & d_{1,16} \\ \vdots & & \ddots & \vdots \\ d_{9,1} & d_{9,2} & \dots & d_{9,16} \end{bmatrix} \cdot \bar{F} \quad (5)$$

90 Muscle dynamics were formulated implicitly through the force balance equation
91 (19):

$$92 \quad F_{SEE}(l_{CE}) = a F_{max} f(l_{CE}) g(\dot{l}_{CE}) + F_{PEE}(l_{CE}) \quad (6)$$

93 Where F_{SEE} is the force in the series elastic element, l_{CE} is the muscle fiber
94 length, a is the activation state, F_{max} is the maximal isometric force, f and g
95 are the normalized force-length and force-velocity relationships, and F_{PEE} is
96 the force in the nonlinear parallel elastic element. Activation was controlled
97 by neural controls u , bounded between zero and one, through an activation
98 dynamics model. Details can be found in (19).

99 The multibody dynamics equations 4 and 5 were combined with the muscle
100 dynamics, equation 6, and the ground contract model to form a set of implicit

101 differential equations, $f(x, \dot{x}, u) = 0$, with $x = [q, \dot{q}, l_{CE}, a]^T$. This function was
 102 coded in C and compiled as a MEX-function for MATLAB (Mathworks, Natick,
 103 MA). The partial derivatives of the function with respect to x , $\dot{x}$ and u were also
 104 calculated by the function, as required for gradient-based optimization (19).

105 2.2. Optimal Control Problem

106 Predictive simulations were performed by solving an optimal control prob-
 107 lem: find periodic state trajectories $x(t)$ and control trajectories $u(t)$ that satisfy
 108 the system dynamics and minimize a cost function. The formulation and so-
 109 lution method were identical to (16), but no left-right symmetry was assumed
 110 and a full gait cycle was simulated. A two-part cost function was used, using a
 111 weight w to determine the relative importance of each objective:

$$112 \min_{\bar{x}, \bar{u}} F_1(\bar{x}, \bar{u}) + w \cdot F_2(\bar{x}, \bar{u}) \quad (7)$$

113 The first objective, F_1 , was used by van den Bogert et al. (16) to perform pre-
 114 dictive simulations of walking and running. This objective penalizes muscular
 115 effort and deviation from normal joint angles, GRFs and gait duration:

$$116 F_1(\bar{x}, \bar{u}) = \frac{1}{11} \left(\frac{1}{T} \sum_{k=1}^{10} \int_0^T \left(\frac{s_k(\bar{x}(t)) - x_{m_k}(t)}{\bar{\sigma}_k} \right)^2 dt \right. \\ 117 \left. + \left(\frac{T - T_m}{\sigma_T} \right)^2 \right) + \frac{W_{\text{effort}}}{16T} \sum_{j=1}^{16} \int_0^T u_j(t)^2 dt \quad (8)$$

118 The eleven tracked variables $s_k(\bar{x}(t))$ were six joint angles, the horizontal and
 119 vertical GRFs of both legs, and the duration of the gait cycle, T . Tracking
 120 data, $x_{m_k}(t)$ and T_m , and standard deviations, $\bar{\sigma}_k$, were provided by Winter
 121 (20). Based on (16), $W_{\text{effort}} = 10$ was used to scale the effort term.

122 The second objective function, F_2 , was added to study the effect of encour-
 123 aging symmetric joint moments in knee and hip. Joint moment symmetry is
 124 encouraged indirectly by the tracking term in F_1 , but the amputation makes
 125 the model asymmetrical. Asymmetrical movement will therefore occur if this
 126 reduces the effort term in F_1 . Furthermore, even if joint angles and GRFs are
 127 symmetric, there may be asymmetry in the center of pressure, which affects joint

128 moments but is not included in the tracking term. Joint moment asymmetry
 129 was defined as the root mean square difference between the left and right joint
 130 moment, accounting for a phase shift of a half-cycle:

$$131 \quad F_2(\bar{x}, \bar{u}) = \frac{1}{2} \sum_{i=1}^2 \int_0^T \left(M_{i_r}(t) - M_{i_l} \left(t + \frac{T}{2} \right) \right)^2 dt \quad (9)$$

132 The optimal control problem was solved through direct collocation, with 60
 133 collocation nodes per gait cycle and a backward Euler formulation. One node
 134 was added to ensure periodicity of the gait cycle, so the final state matches the
 135 initial state with a forward translation. Details can be found in (15) and (16).
 136 The resulting constrained optimization problem was solved using IPOPT 3.11.0
 137 (21).

138 First, the optimal control problem was solved for the able-bodied model
 139 (ABLE). To obtain this solution, standing was solved first, followed by a series
 140 of able-bodied gait solutions with increasing speed. Then, using the able-bodied
 141 solution as an initial guess, TTA gait was solved using only objective F_1 , thus
 142 without considering asymmetry in joint moments (TTA1).

143 Then, objective F_2 was added, creating a multi-objective optimization prob-
 144 lem for TTA gait due to the asymmetry between the left and right leg. A
 145 Pareto-front was obtained by solving the optimization problem using different
 146 weights for objective F_2 .

147 Two solutions (TTA2 and TTA3) on the Pareto-front were analyzed and
 148 compared against the baseline and the able-bodied solution to analyze the effect
 149 of F_2 . Joint angles, joint moments, muscle forces, GRFs and joint contact forces
 150 were examined to find kinetic and kinematic changes. The energy required for
 151 each of the predicted gaits is compared using the effort term in equation 8 and
 152 a metabolic cost model (22). GRFs, joint angles and muscle forces were used
 153 to find the joint contact force (JCF) magnitudes, the Euclidean norm of the
 154 components in the x- and y-direction.

3. Results

First, able-bodied gait was solved. A gait cycle of 1.14 seconds with a speed of 1.325 m/s required 3.12 W/kg metabolic energy. Joint angles and ground reaction forces of this solution were all within one standard deviation of the tracking data (20). Only the peak hip flexion angle was slightly higher. The able-bodied solution was found in 7 minutes and 52 seconds on a computer with an Intel Core i5-3210M CPU at 2.5 GHz clock speed.

The Pareto-front (Figure 2) shows the trade-off between symmetry in the joint moment and the objective of low effort and normal gait. Initially, the increase in joint moment symmetry had a small effect on the tracking and effort term. However, as joint moment symmetry improved, the cost of more joint moment symmetry increased considerably. All gaits are theoretically feasible for a person with a TTA, the choice between them depending on the relative weight of the optimization objectives, which are reported in figure 3.

Three points are of interest in the Pareto-front. Solution TTA1 is the baseline solution, where the weight of objective F_2 was zero. Solution TTA2, which used $w = 0.3$, is a solution where the symmetry of the joint moments greatly improved with little extra effort and deviation from normal kinematics (11% increase in F_1). Solution TTA3, where $w = 10$, required an increase in effort and tracking error of almost 33%.

Figure 3 shows the tracking error, effort, RMS moment asymmetry and metabolic cost as a function of the weight w in equation 7. Tracking error increased almost linearly from 0.55 at $w = 0$ to 0.79 at $w = 10$. Effort stayed approximately constant at 0.0257 until $w = 0.2$, after which it increased to 0.0282 at TTA3. Joint moment asymmetry decreased from 17.5 Nm to 5.19 Nm at $w = 0.2$ and then to 0.635 Nm at solution TTA3. The metabolic cost first decreased from 3.09 W/kg to 3.05 W/kg at $w = 0.05$, after which it increases to 3.21 W/kg at TTA3.

Joint angles (Figure 4) were within one standard deviation of the tracking data, except the peak hip flexion angle at heel strike, which deviated more

185 than two standard deviations, and the intact knee angle in TTA3 between 20%
 186 and 40% of the gait cycle, which deviated more than one standard deviation.
 187 Furthermore, the knee and hip joint kinematics were similar and nearly sym-
 188 metric in all solutions. However, the joint kinetics distinctly differed between
 189 the solutions.

190 In TTA 1, which did not consider joint moment asymmetry, joint moments
 191 were distinctly asymmetric and differed from ABLE (Figure 4). The peak ex-
 192 tension moment in the hip of the prosthesis side was 38 Nm, compared to 100
 193 Nm on the intact side and 44 Nm in ABLE. The peak extension moment in the
 194 knee was 28 Nm on the prosthesis side and 51 Nm in the intact side, compared
 195 to 42 Nm in ABLE.

196 In solution TTA2, tracking error increased by 16%, effort increased by 2%
 197 and joint moment asymmetry was reduced by 76%. Peak extension moment in
 198 the hip was 67 Nm on the prosthesis side and 76 Nm on the intact side, more
 199 symmetrical than in TTA1 but higher than in ABLE. The peak knee extension
 200 moment was also more symmetrical, 26 Nm on the prosthesis side and 34 Nm
 201 on the intact side. Both values were lower than in ABLE.

202 Solution TTA3 reduced joint moment asymmetry by 88%, but required an
 203 increase in effort of 10% and in tracking error of 43%. In this solution, the peak
 204 hip extension moment was equal to 89 Nm on the prosthesis side and 90 Nm on
 205 the intact side, more than twice as high as in ABLE. The peak knee moment is
 206 equal to 22 Nm on the prosthesis side and 23 Nm on the intact side, which is
 207 about half that of ABLE.

208 Figure 5 shows the muscle forces in the musculoskeletal model. Hip muscle
 209 forces were lowest in ABLE. In TTA1, the Hamstrings force was high in early
 210 stance on the intact side, 603 N, compared to 356 N on the prosthesis side.
 211 The peak force in the Gluteals was 986 N on the intact side, compared to 307
 212 N on the prosthesis side. Also, the peak force in the Iliopsoas was high on
 213 the prosthesis side during late stance, 883 N, compared to 280 N on the intact
 214 side. In TTA2 and TTA3, peak forces in the hip muscles on the prosthesis side
 215 increased compared to TTA1. The peak force in the Gluteals on the intact side

216 decreased, and so did the peak force in the Hamstrings in TTA2. The other
217 peak forces remained similar to TTA1.

218 The vertical GRF on the prosthesis side was similar for all solutions and
219 within one standard deviation of the tracking data (Figure 6). Large differences
220 were seen in the JCFs (Figure 6). Hip JCFs were higher than in ABLE for both
221 legs in all conditions. In the intact side, the peak JCF was lowest in TTA2,
222 but higher than in ABLE. The JCF in the knee of the prosthesis side was lower
223 than ABLE for all solutions.

224 4. Discussion

225 An able-bodied solution was found using the approach presented by van den
226 Bogert et al. (16). Joint moments are symmetric between the two legs. Their
227 values correspond to results found in studies with human subjects. Also, the
228 metabolic energy was similar to metabolic energy found in studies with human
229 subjects (23).

230 Any of the solutions on the Pareto-front can theoretically be chosen as a gait
231 by persons with a TTA. Therefore, the hypothesis that persons with a TTA
232 can walk with more symmetry in the joint moments at the cost of increased
233 effort and abnormal kinematics is confirmed. Figure 3 shows that joint moment
234 asymmetry can be reduced by 70% with little increase in effort or metabolic
235 cost.

236 If persons with a TTA select a gait with normal kinematics and minimal
237 effort, we would expect them to select solution TTA1. The largely symmetric
238 joint angles in the knee and hip in TTA1 are also seen in studies of persons
239 with a TTA (11; 24). The predicted reduction of the knee joint moment on the
240 prosthesis side in TTA1 is also seen in subjects with a TTA (8; 11; 12).

241 Nolan and Lees (25) and Ferris et al. (8) report a higher peak extension hip
242 moment in the intact leg compared to able bodied walking, which agrees with
243 our predictions in TTA1. However, some literature reports more symmetry in
244 the peak hip extension moment of the intact and residual leg (26; 27), which is

245 more similar to TTA2. TTA1 and TTA2 require a similar amount of effort or
246 metabolic work, which suggests persons with a TTA operate around TTA1 and
247 TTA2 in the Pareto-front.

248 Solution TTA2 has several advantages over TTA1 and TTA3. A lower peak
249 extension moment in the hip and knee reduced joint moment asymmetry and
250 yielded a lower JCF in both joints. A high JCF is suspected to be related
251 to a higher incidence rate of osteoarthritis in persons with a TTA (2), so the
252 lower JCFs could decrease this higher incidence rate. Another potential benefit
253 in TTA2 is the higher knee JCF in the residual leg compared to TTA1 and
254 to ABLE, which might reduce the loss of bone mineral density in the proximal
255 tibia and femur neck, which is reported for persons with a TTA (28). Therefore,
256 teaching persons with a TTA to walk using TTA2 may be useful in rehabilita-
257 tion.

258 TTA3 emphasized joint moment symmetry too much, which is disadvanta-
259 geous for TTA gait. Apart from a 10% increase in effort compared to TTA1,
260 the peak hip extension moment increased in both legs, which also yielded a
261 higher JCF in the hip. Currently, prevalence of osteoarthritis in the intact hip
262 is already higher in persons with a TTA (3) and TTA3 could increase the risk
263 of developing osteoarthritis in the hips of both legs.

264 On the prosthesis side, there is no active push-off and the biarticular Gastroc-
265 nemius is removed completely. This might cause the Iliopsoas force to increase
266 on the prosthesis side during late stance, because this muscle now initiates the
267 swing phase and compensates for the activation of the biarticular Hamstrings,
268 which flexes the knee instead of the Gastrocnemius. Some persons with a TTA
269 can still use the Gastrocnemius to control the knee, which might reduce this
270 effect. Additionally, the lack of push-off increases the impact force on the in-
271 tact side (Figure 6) (29). A passive foot that stores and releases energy using
272 microprocessor control, such as the artificial foot by Collins and Kuo (29) or a
273 prosthesis with a motor can better restore the push-off function of the biological
274 ankle and reduce these effects.

275 The metabolic model produced a lower metabolic cost for TTA gait than

for able-bodied gait, which contradicts subject studies (23). The predicted metabolic cost for TTA gait is lower because three muscles are removed that would otherwise contribute significantly. While our predictions are theoretically possible, actual TTA gait may require more energy due to antagonistic cocontraction to stabilize the prosthetic side (6), for example due to an unstable connection between the prosthesis and residual leg. Such co-contraction will not be predicted by a simulation where effort is optimized and no stability is required. Therefore, we conclude that other objectives might also influence TTA gait, like stability, joint pain or comfort. The underestimation of metabolic energy suggest that we may also have underestimated the muscle forces and JCFs.

The modeled prosthesis had the same mass and moment of inertia as the intact leg. This ensured that this study only predicted effects due of control and not of a change in prosthesis parameters. Prosthetic legs have different mass and inertial properties than a normal leg, but this will mostly affect the swing phase, where forces and moments are low. Therefore, we expect that there is little effect of mass properties on the peak loading variables.

This study used a two-dimensional musculoskeletal model. Hip joint forces may have been underestimated by neglecting the abductor moments. However, we expect the main conclusions to be valid. This should be verified with a three dimensional musculoskeletal model or with subject tests. Further tests with a three dimensional model could also analyze the effects of having joint moment symmetry in the frontal and transverse plane, while a subject study could verify if subjects can maintain the suggested gait modifications.

We conclude from this study that:

1. Asymmetric knee and hip joint moments in persons with a TTA with a passive foot can be explained by a predictive simulation model that minimizes effort and deviation from normal kinematics.
2. Considerable improvement in moment symmetry is theoretically possible with a small increase in effort and almost no change in kinematics. The

306 hip joint moment will, however, remain elevated above able-bodied values,
307 which is currently also reported for persons with a TTA (26; 27).

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312 6. Conflict of Interest Statement

313 There are no conflicts of interest for any of the authors regarding the research
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315 7. References

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Figure 1: Sagittal plane model that is used in the predictive simulations.

Figure 2: Pareto-front showing the trade-off between effort and symmetry in the joint moments. Solutions TTA1, TTA2 and TTA3 will be analysed further.

Figure 3: Figure showing the different objectives of tracking error, effort and moment symmetry, as well as the metabolic cost as a function of the weight w .

Figure 4: Joint angles, moments for the prosthesis side (left) and the intact side (right) for solutions ABLE, TTA1, TTA2 and TTA3. The gait cycle starts at heel strike on both sides. The angles show flexion positive, while in the moment graphs, extension is positive.

Figure 5: Muscle forces of all muscles in the musculoskeletal model for solutions ABE, TTA1, TTA2 and TTA3. The gait cycle starts at heel strike on both the prosthesis side (left) and the intact side (ride).

Figure 6: Joint contact force in the knee and hip for solutions ABLE, TTA1, TTA2 and TTA3.