Development of dynamic models of the Mauch prosthetic knee for prospective gait simulation

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Abstract

Recent advances in computational modeling and simulation of human movement make it possible to isolate and predict the potential contributions of a prosthetic device to the overall system performance. The Mauch S-N-S knee is one of the most widely used prosthetic knees in the market. The goal of this study is to develop dynamic models of the Mauch S-N-S knee for predictive simulation of a transfemoral amputee’s gait under idealized conditions. Based on the functional description of the Mauch S-N-S prosthetic knee from the literature, a combined bench test and data fitting approach employing modified slow, normal, and fast gait patterns and nine combinations of stance and swing damping settings were performed. Two types of dynamic models, 2-phase and 4-phase models, of the Mauch S-N-S prosthetic knee were developed. The range of the coefficient of determination of the two dynamic models, when compared to the test data, was from 39.9 to 95%. Both dynamic models of this study can be utilized in musculoskeletal modeling studies, to better understand amputee gait and the contributions and interactions of various prosthetic leg components to the ambulatory performance.

1. Introduction

Transfemoral amputation is a disabling condition that significantly impacts patient mobility and quality of life (Basu et al., 2008). As far back as the ancient Egyptian dynasties, attempts were made to improve prosthetic legs to permit more natural gait (Norton, 2007). Prosthetic knee innovation often followed an increase in amputations due to warfare (Vanderwerker, 1976). Development of the “Mauch” knee, available from Ossur Hf. (Reykjavik, Iceland) began after the Second World War (Mauch, 1958, 1968). Since then, prosthetic knees of increasing sophistication have been developed with the expectation of superior performance (Bunce and Breakey, 2007; Orendurff et al., 2006). Walking with a prosthetic leg limits the quality of life of most transfemoral patients and health problems consequent to sedentary lifestyles are common (Akarsu et al., 2012; Ehde et al., 2001; Giummara and Bradshaw, 2010). Are newer and better knees needed, or are current knee prostheses’ capabilities underutilized because of other issues with amputees or other prosthetic leg components? Answers to these questions may be difficult to find solely by human-subjects testing, which include many confounding factors, e.g., prosthesis fit, socket performance, and patient rehabilitation/motivation (Jin et al., 2003). Experimentally, it is difficult to isolate the effects of the prosthetic knee on amputees’ gait. Computational modeling of a prosthetic leg in the context of gait simulation studies (Ackermann and van den Bogert, 2010), can offer a cost-effective and systematic methodology to isolate and evaluate the capacity of any given component of the prosthetic system to reproduce ideal gait. Improvements in the configuration...
of prosthetic legs and leg components, all targeted at improving the capability of the component to achieve a desirable kinematic-kinetic capacity can be evaluated. Examples of computational approaches to studying amputee gait include transfemoral amputee gait (Strbac and Popovic, 2012; Fey et al., 2012; Au and Herr, 2008) and transfemoral amputee gait (van den Bogert et al., 2012). With the development and application of dynamic models, it becomes possible to determine the optimal internal features of each prosthetic component to result in optimal performance of the system.

Our goal is to establish a mathematical model quantitating the functional capacity of the widely utilized Mauch Knee, as a precedent for functional evaluation of future prosthetic knee designs in silico. The specific goals of this study are to (1) acquire experimental data establishing the kinetic-kinematic response of a Mauch Knee over a range of gait conditions and choices for the damping settings built into this design, and (2) develop dynamic models of the Mauch Knee that can be used in movement simulations.

2. Methods

A previously used “Mauch Gaitmaster Low Profile SNS Jr.” (Ossur, Reykjavik, Iceland) knee cylinder was tested on the bench, setting its performance adjustments at various levels. The hydraulic cylinder was driven at various kinematic profiles representing gait at various speeds while measuring force and cylinder position. Multiple regression techniques were then applied to estimate coefficients for dynamic models ("2-phase" and "4-phase", as described below) of the prosthetic knee based on data from walking at a normal speed. The dynamic equations applied to slow and fast walk data to evaluate the effectiveness of the models at speed conditions different than those used to derive the equations.

2.1. Test sample

History of the tested Mauch Knee is unknown but the device showed no sign of leaks or abuse and a veteran orthotics technician examined and judged it to be in normal working order. Hans Mauch has described the construction and function of the design (Mauch, 1968), which is summarized here. The device consists of a hydraulic damper that provides high resistance to knee flexion during stance phase and lower resistance during swing phase. The user can access two adjusting dials to separately modify knee flexion and knee extension damping. These dial adjustments are referred in our study as “E” for extension and “F” for flexion. Dial settings from low, medium to high in each case are designated as 0°, 90°, and 180° with respect to the dial position in this research. The Mauch Knee incorporates a hydraulic cylinder, with a valve and balance wheel-based mechanism that sets the knee state as swing or stance, and flexion or extension. This arrangement is sensitive to flow direction, viscosity, inertia, gravity and knee hyperextension. During knee flexion in stance phase, the valve closes the main orifices and the hydraulic fluid can only flow through small cut-outs around the piston which results in high resistance. When the knee hyperextends, the piston lifts a pawl, which unlocks the balance wheel and if this hyperextension condition persists for a sufficiently long duration, i.e., 1/10 s in Mauch’s estimation (Mauch, 1968), the balance wheel will rotate far enough to prevent valve closure. This situation is most probable at late stance; hyperextension at end swing is possible but is not likely to persist long enough for the mechanism to act; the piston will retract and the pawl will drop before the balance wheel rotates far enough to be free. Balance wheel inertia and viscous drag establish the delay time. High extension flows through the valve orifice maintain valve stem pressure against the balance wheel and prevent it from rotating back into the locked position until extension flow through the valve stops and the valve poppet drops.

2.2. Bench testing

To develop a dynamic model, extensive data sets matching input conditions, e.g., hydraulic cylinder displacement trajectories, and output results, e.g., hydraulic cylinder force trajectories, are required. To ensure that the data collection covered a range of variables relevant to gait, slow, normal, and fast walking data from literature, specifically knee angle trajectories, were used (Winter, 2005). The prosthetic knee geometry was utilized to calculate cylinder position (BC) vs. time profiles using these data (Eq. 1, Fig. 1)

\[ BC = \frac{m^2}{C0} + \frac{m^2}{C2} - 2m \times \frac{m^2}{C0} \times \cos(\alpha) \]

where \( \alpha=90° \) – (knee flexion angle), \( \frac{m^2}{C0} = 0.1603 \) m, \( \frac{m^2}{C2} = 0.0247 \) m
The knee angle profiles and the associated calculated cylinder displacement profiles (Fig. 2) were slightly modified to take into account the operating mode (as described above) of the Mauch Knee. The minimum knee angle at the end of knee extension during stance phase was shifted down relative to the original profile (Fig. 2A) such that it enters the hyperextension position while the total angular displacement to the peak angles of stance and swing phase remained the same. Fig. 2B represents the conversion of the modified knee angle into linear displacement of the hydraulic piston. Four time points before and after the modified minimum knee angle in the end knee extension during stance phase were made equal to the modified minimum knee angle at end knee extension during stance phase to ensure sufficient duration of this position (Fig. 2C). The goal of this test is to collect knee

![Fig. 1. General geometry of a prosthetic knee.](image-url)
mechanism data over a wide range of conditions and the test points need not match any particular subject’s gait so long as the knee function is fully exercised. Conversion of the angular knee motion to linear motion of the cylinder was made using Eq. (1).

The cylinder was removed from the knee and mounted into a testing machine (MTS 858 Bionix, Eden Prairie, MN, USA) (Fig. 3). The testing setup applied the piston displacements while recording the resulting displacements and forces simultaneously. No discrepancy was shown between the desired and actual piston displacement. The data were collected at 361.4 Hz, which ensured acquisition of more than 300 data points for a single gait cycle. Each test gait profile was cycled 5 times to demonstrate a reproducibility of the input–output data pairs. The testing was repeated for nine combinations of flexion and extension damping values, specifically 0, 90, and 180, and three walking cadences, specifically slow, normal, and fast. A total of 9 nine dial conditions for 3 walking speeds resulted in 27 test conditions.

2.3. Model conceptualization

Two dynamic models were developed, based on the knowledge of the Mauch Knee mechanics and the test data. Each model was formulated to be differentiable, which is required by the mathematical solution algorithms of simulation methods (Ackermann and van den Bogert, 2010).

The 2-phase model is a simplified representation of the hydraulic system of the Mauch S-N-S knee. The experimental force-displacement response of the Mauch Knee cylinder suggested two basic conditions (Fig. 4A): Phase 1, the high force section (at stance flexion); and Phase 2, the low force section (all other gait phases). The governing force equation of the piston of the 2-phase model was written as

$$F_{pis} = (1 - S_3) \left( c_1 \dot{x} + k_1 x + f_1 \right) + S_3 \left( c_2 \dot{x} + k_2 x + f_2 \right)$$

where $x$ and $\dot{x}$ are the piston position and velocity, respectively, $c_1$, $k_1$, $f_1$, $c_2$, $k_2$, and $f_2$ are coefficients of the dynamic equation to be determined for each phase. $S_3$ is a phase switch variable (Eq. (3)) which selects for the equation to be used at a given time. It is determined by two subordinate switch variables $S_1$ and $S_2$. $S_1$ establishes the extension/flexion state of the knee (Eq. (4)) from the knee angle velocity. $S_2$ indicates whether or not the knee has been in the hyperextension state (Eq. (5)).

$$S_3 = \frac{1}{C_0} \left( \frac{1}{C_1} + \frac{1}{C_2} \right) \tanh \left( \frac{\dot{x}}{x_0} \right)$$

$$S_1 = 1 - S_3 (1 - S_2)$$

$$S_2 = \int S_2 \, ds$$

Fig. 2. (A) Solid line, the original knee angle profile during normal walking. Dashed line, modified knee angle profile to ensure entering the hyperextension position after mid-stance phase. (B) Dashed line, conversion of the modified knee angle into linear displacement of the hydraulic piston. Solid line, extended period of time in the hyperextension position to ensure triggering of hyperextension mode. (C) The extended piston displacement within the circle region of (B).

Fig. 3. Experimental setup to characterize the kinematic-kinetic response of the Mauch S-N-S prosthetic knee cylinder on MTS 858 Bionix test system (MTS, Eden Prairie, MN USA). Within the circle is the hydraulic cylinder of the Mauch S-N-S prosthetic knee.
where $x_{hy}$ is the hyperextension position (determined experimentally to be 0.005 cm), and $k$ is another constant controlling the rate at which the parameter changes. A value of $k = 10,000$ was chosen for the same considerations as governed $S_1$, a rapid transition without significant non-linearity.

The 4-phase model is a more complex model of the hydraulic system which incorporates design features of the knee as well as the test data. As constructed, the Mauch Knee operates in different phases to provide appropriate resistance through the gait cycle (Fig. 4B): Phase 1 represents the knee flexion in early stance phase, where high resistance prevents knee collapse; Phase 2 represents knee flexion at end stance and early swing phase, where knee resistance is moderate; Phase 3 represents knee extension in both stance and swing phase, where low resistance eases knee extension; and Phase 4 represents the hyperextension mode. The governing force equation of the piston of the 4-phase model is

$$F_{pis} = F_1 + F_2 + F_3 + F_4$$ (7)

where $F_1$, $F_2$, $F_3$, and $F_4$ are the force equations of phases 1, 2, 3, and 4, respectively

$$F_1 = S_4(c_1 x + k_1 x + f_3)$$ (8)

$$F_2 = S_5(c_2 x + k_2 x + f_4)$$ (9)
where $c_{1..4}$ and $f_{1..4}$ are the coefficients applicable to each phase equations. $S_{1..7}$ are phase switch variables, which dictate the utilization of a specific equation. These switches are functions of the flexion/extension state, the initiation of the hyperextension state, and the continued presence of hyperextension. Useful values of the four operators can be developed using the previous equations for $S_1$ and $S_2$, along with a new term, $S_8$, which establishes the initiation of the hyperextension mode

\begin{align}
S_4 &= S_1 \times (1 - S_2) \times S_8 \\
S_5 &= (1 - S_1) \times S_8 \\
S_6 &= S_1 \times S_2 \times S_8 \\
S_7 &= 1 - S_8 \\
S_8 &= \frac{1}{2} \left[ \frac{1}{2} \tanh \left( \frac{x - x_0}{x_{hy}} \right) \right]
\end{align}

### 2.4. Parameter estimation

Given the kinetic-kinematic response of the Mauch Knee hydraulic system, the data were divided into groups associated with each phase of the model under development. Multiple linear regression was utilized to estimate coefficients for force equations ($c$, $k$, and $f$) for each model phase. These coefficient estimations minimized the difference between predicted cylinder forces and those measured. Only normal walking data were used for data fitting and the process was repeated for each of the Mauch Knee dial settings. The slow and fast data were reserved to evaluate the ability of the model to predict performance at conditions distant from those used to derive it.

### 2.5. Model evaluation

Following the model coefficient determination, the performance of the dynamic models was tested against the experimental slow and fast walk data. The models’ resulting performance for different data sets was evaluated statistically by calculating the maximum error (ME), the root-mean-square error (RMSE) and the coefficient of determination ($R^2$) between measured and predicted cylinder force. These three methods could provide information relating to worst prediction, relative prediction errors, and average prediction of the dynamic models.

### 3. Results

The coefficients for the 2-phase and the 4-phase models obtained from normal walking data at external dial settings of F90E90 are provided in Table 1. The fit errors for these models are summarized in Table 2. Performance of the model for the same dial setting at different walking speeds can also be found in Table 2. Model coefficients for the 2-phase and the 4-phase models are shown in Appendices A and B, for 2-phase and 4-phase models, respectively. Model evaluation metrics for different speeds of walking can be found in Appendices C and D, for 2-phase and 4-phase models, respectively.

Predicted forces for dial setting F90E90, as obtained from 2-phase and 4-phase models, are shown in Fig. 6. The functioning of the switch variables in the 2- and 4-phase models is shown during the modified normal gait pattern on Fig. 5.
4. Discussion

In this study, dynamic models of the Mauch Knee hydraulic cylinder were developed using normal walking experimental data to identify the coefficients for the models, and slow and fast walking data were used to validate the predictions of the resulting models. The models incorporate damping, stiffness and a constant offset force. The damping and stiffness are built-in functions of the mechanism as described by Mauch. The offset force is not discussed by the designer, but it appears to be a preload force built into the system. Modeling as a friction that consistently opposed the direction of motion resulted in a less effective model. Including an acceleration term did not improve the model, probably because the piston and rod inertia was low, but did complicate the equations.

For both models, the maximum error occurred in the transition of knee extension to hyperextension in all dial settings. This is attributed to phase switching in the models not exactly matching the experimental behavior, because the timing of the releasing and rotating the balance wheel resulting from an interaction of viscous and inertia effects is not well-duplicated in the model. The constants $k$ and $x_0$ in this study were selected to be 0.05 and 10,000 to cause a rapid yet continuous phase switch as discontinuities in numerical representation of the Mauch Knee system can be expected to cause convergence problems for gait simulation studies. Conceptually, a more complex system identification method that included these switching parameters as additional coefficients to be numerically determined might have identified better values than the choices made here (Ljung, 1999).

As expected, the 2-phase model had a lower accuracy (Table 2) since certain mechanical aspects of the internal design were not incorporated. Representation of the high force region, when the internal valve is closed, was more faithful to the data. The low force region incorporated a variety of conditions without differentiation and therefore the model should be expected to have less accurate predicted results. The 4-phase model incorporated had a finer resolution, including more switch parameters to differentiate phases and represent transitional behavior. In return for the complexity, a more accurate model was obtained.

Ideally, the derived model should accurately predict the experimental data at all walking speeds. However, as expected, the estimated models performed better when compared to the normal gait data from which the model coefficients were derived (Table 2, Fig. 6) and the 4-phase model did better than the 2-phase model when compared to experimental data. The worst model prediction is for the slow gait of the 2-phase model which the $R^2$ value went as low as 49.1%. Performance degraded at the other conditions, particularly for slow gait. Both dynamic models assume linear behavior of the mechanism for a given phase, and the knee dynamics during slow gait may have been more influenced by non-linear hydraulic fluid flow effects in the cylinder, orifices, and valve due to slower flow. Yet, this discrepancy (and associated inaccuracies) may be acceptable for the desired utility of these models in the future, i.e., for simulation of gait. A larger data set which allowed higher and slower speed data to be used for coefficient estimation while still reserving some independent data for evaluation might have resulted in blending the errors and resulting in better overall prediction. However, if fluid mechanics or other effects are changing the fundamental mechanics of the design as speed changes, a more complex system model may be necessary for significantly better results. The 2-phase model is less accurate but simpler than the 4-phase model and may be easier to use for a complex system simulation. Furthermore, it can be observed that the hyperextension mode (phase 4 in 4-phase model) of the system is where the 2-phase model has the largest deviation of force prediction. If the hyperextension mode does not occur in the prospective simulations, the adequacy of the 2-phase model is further enhanced. More complex models than those proposed here are possible, but the effort to obtain sufficient experimental data and achieve convergence of the regressions may be significant, while the resulting more involved models may cause computational issues for the next stage simulations without dramatically improving simulation results.

The models were derived for nine sets of damping settings. Performance of the modeling process was similar for all models. As damping increased, the coefficient of determination improved across the models.

This analysis applies strictly to a single used Mauch Knee cylinder. When properly integrated into the geometry of a complete prosthetic knee we believe it will yield a useful model for further integration into a human gait simulation algorithm. The purpose of using the modified gait pattern as test input is to operate the Mauch Knee with a reasonable gait pattern while ensuring that all the prosthetic knee functionalities were exercised. No single subject is likely to walk with these gait patterns, but the model has been evaluated over a sufficiently broad range that we expect that most patterns will fall within the testing.

The dynamic modeling methodology employed was effective and straightforward; the results apply strictly only to the Mauch knee that has been tested in this study. It was based on understanding the design function and operational data. A similar approach can potentially be applied to other Mauch Knee samples, or to other artificial knee designs, including microprocessor-controlled designs (Kirker et al., 1996; Orendurff et al., 2006), and to other prosthetic components, e.g., the prosthetic ankle/foot. In each case the variables of the model will have to be chosen to suit the hardware under test. The switching between operational phases would be based on the specific design logic of the hardware under consideration.

For transfemoral amputee gait simulation, this type of dynamic model can be used to estimate the kinematic/kinetic performance of the prosthetic knee when incorporated into the musculoskeletal model. The representative anatomic joint features would be removed in the musculoskeletal model after a “transfemoral amputation” and the knee model equations would be added into the mathematical system of the musculoskeletal model. With proper models of the other components, it should be possible to build system level models of a complete prosthetic leg and subject which could be utilized to predict the functional capacity of a given prosthetic leg, and to compare this predicted functionality to the published gait kinetics and kinematics data of patients who use the prosthetic hardware under study, models of other hardware, or other available test data. This will enable (1) compartmentalization of the gait limitations of each modeled subsystem under the assumed subject conditions, (2) new insights for better component design, selection, integration and, in the long run, improved patient rehabilitation, and (3) the need for exploratory field testing with patients to develop and refine prostheses may be reduced.

Conflict of interest statement

None.

Acknowledgments

The authors acknowledge the contribution of the State of Ohio, Department of Development and Third Frontier Commission (TECH 09-001), which provided funding in support of this
research. The authors would also like to thank Kirsten Richard and her associates at the Orthopedic & Rheumatologic Institute of Cleveland Clinic for technical support and inspection of the state of the test sample.

Appendix A. Supporting information

Supplementary data associated with this article can be found in this online version at http://dx.doi.org/10.1016/j.jbiomech.2014.06.011.

References