DESIGN & DEVELOPMENT

1990

"DUPLICATING HYDRAULIC GAIT"

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Abstract

Previous research studies indicated that the pendular characteristics of the prosthetic shank could be manipulated to achieve an increased walking velocity. This project looked at the possibility of duplicating the gait achieved with a Hydraulic knee mechanism at three different walking speeds with a Constant Friction knee unit through changing the pendular characteristics of its shank. Results showed that duplication was not possible with identical shank masses yet was very successful with a greater shank mass. Oxygen consumption was tested to evaluate energy expenditure and showed the same gait with both knee mechanisms required very similar energy output.
A dilemma which faces the Above Knee Amputee and Prosthetist is often the type of knee unit to be used. In many cases a patient who may be a perfect candidate for a Hydraulic Knee unit often goes without due to the expense of such a unit. This research project looked at the possibility of adapting a Constant Friction knee unit, a cheaper and simpler knee unit, through the addition of weight, to give a gait pattern equal to that of an expensive Hydraulic unit.

The matter of ideal prosthesis weight has yet to be agreed upon.

Van De Veen (1957) developed a mathematical model of an Above Knee prosthesis to investigate the influence of mass application and mass distribution on stump load and the energy required for ambulation. It was found that energy output increased with the addition of weight yet the degree of output was dependant on the distribution of the weight.

Godfrey (1977) examined a series of patients to determine the effect on gait when the mass of the prosthesis was increased. Using weights of 113 and 227 grams it was found that gait characteristics were unchanged with the addition of mass.

Inman (1967) concluded that the ideal Above Knee Prosthesis would be light at the beginning of swing requiring little energy to initiate swing and heavy at the end of swing providing enough kinetic energy to be fed back into the body to maintain forward movement.

It is agreed that Hydraulic knee units are heavier than Constant Friction knee units and possibly coincidentally, this weight difference helps provide a more symmetrical gait.

One may question the contribution of the mass of the Hydraulic unit in achieving its gait pattern.

The concept of investigating the chosen subject came as a consequence of reading primarily three different references. An explanation of these references is required prior to a detailed description of this investigation.

Murray (1963) found that Above Knee amputees regardless of knee type, Hydraulic (H) or Constant Friction (CF), walk with a slower cadence than non amputee controls. Walking at differing speeds it was found that a "H" knee gave a more symmetrical gait and a cadence more similar although considerably slower, to the normal than did the "CF" knee. It was found that a "H" knee achieves a faster walking velocity through achieving a faster cadence rather than an increase in stride length. The slower cadence of the "CF" knee was attributed to a too long a swing phase and this was in turn attributed to an excessive amount of heel rise via excessive knee flexion after toe off. Another aspect of the "H" knee as compared to the "CF" knee was its ability to allow the amputee to walk at a wider range of speeds. We therefore see that the main downfall for the "CF" knee is that it has too long a swing phase. A problem soon found capable of correction.

Tashman, Hicks and Jendrzejczyk (1985) based a research project on an investigation carried out at the Newington Children's Hospital Kinesiology Laboratory. Using "CF" knees, this investigation found that a prosthetic shank through its own pendular characteristics has a predicted natural swing period. This period is slower than that of the non-amputated shank, hence the slower cadence of Above Knee amputees. An amputee tends to have a comfortable walking speed which is predicted by the natural swing period of the shank. This walking speed remains that of the natural swing period of the shank because of the considerable amount of energy required to force the shank to swing faster.

Tashman et al (1985) set out to increase the comfortable walking speed of an amputee via altering the natural swing period of the shank. It was found that by shifting the Centre of Gravity of the shank from normally 31.0cm distal to the knee centre to a point 18.7cm distal to the knee centre, the swing time for the shank was reduced by 5% for both comfortable and fast walking. This resulted in a decreased prosthetic...
Skinner (1969) carried out work on the optimum mass and mass location of the prosthetic shank in an Above Knee prosthesis. Beginning with a very light shank, patients complained that it slowed them down and they had to wait for the foot to swing through. It was found that for comfortable walking, a mass of 1.70 Kg placed either 17 or 25cm distal to the knee centre achieved the most symmetrical gait and was also the most energy efficient. From this reference, it was especially interesting to note that the optimum position of the mass roughly corresponds to the position of a H unit.

These three research programs created an interesting situation when all results were collectively thought through. That is one reference showed that the optimum position for a shank mass was roughly in the same position as a H unit. Another showed that a H unit was superior to a CF unit because it allowed faster walking leading to a more symmetrical gait and also a variety of walking speeds. Thirdly, we saw that "CF" walking speed and hence gait symmetry could be increased by the application of a mass in roughly the area of the position of a H unit.

The three references combined to leave one hypothesizing that the gait symmetry and walking speeds of a H unit may be duplicated with a CF unit via manipulating the centre of gravity of the prosthetic shank. A possibility which is theoretically viable through calculating the Radius of Gyration (distance from knee centre to centre of gravity) for the Hydraulic leg walked at different speeds and simply duplicating this Radius of Gyration for the CF leg. This duplication being achieved by placing a mass equal to the mass difference of the two knee units, (thus having both shanks of identical mass) in strategic positions on the prosthetic shank to achieve identical figures for the Radius of Gyration for the three speeds.

Investigations were carried out in the Biomechanics Laboratory, Lincoln Institute of Health Sciences - La Trobe University.

The investigation found that the duplication of "H" gait with a "CF" knee unit was not possible with the addition of a mass equal to the mass difference of the two knee units directly indicating that the position of the H units mass is not responsible for the gait symmetry achieved. However, gait duplication was extremely successful with a mass greater than the mass difference of the two knees. A mass of 790 grams placed in pre-calculated positions gave the CF knee a gait pattern as good, if not better than the H knee. Assessment of energy efficiency showed little difference in output to walk either the H shank or the Adapted shank.

Results indicating that duplication of all aspects of gait achieved with a H knee unit is definitely possible with a CF knee through the manipulation of the C.O.G. of the prosthetic shank section.
THEORY

To comprehend the basis of this research it is essential to understand why a 'H' knee is superior to a 'CT' Knee

The \textit{H} Knee's units superiority is due to the fact that it has periods of high friction and low friction in the gait cycle and through the use of fluids allows a wide range of walking speeds. Importantly, a 'H' Knee unit is cadence responsive. A 'H' unit utilizes the properties of fluids and fluid flow. The most exploited properties of fluid are, firstly, its tendency to flow with little resistance at low velocities and high resistance at high velocities. Secondly is a fluid's incompressibility or its characteristic that pressure at one point in a fluid is instantly transmitted throughout the rest of the fluid.

One may question the application of these properties to a prosthetic knee unit.

Essentially, knee flexion causes a piston to apply force to a fluid, which is usually an oil. This force is duplicated throughout the fluid, therefore the greater force applied by the piston, the greater force applied to the fluid chamber by the fluid and hence an increase in resistance. The fluid is forced through a set of orifices (only one orifice in the Dynaplex 'H' unit used in this project) the more open orifices leads to lower resistance to flow, thus low resistance to piston movement and vice versa for less orifices. In the case of the Mauch SNS a series of orifices helically line an inner chamber, these orifices provide a pathway for the fluid, which upon compressive forces from the piston can travel to the opposite side of the piston ready for the piston to move in the opposite direction where the same process is repeated. The descending or ascending of the piston sequentially closes these orifices and hence at extremes of piston movement few orifices are open thus great resistance to movement as the fluid is forced through few orifices. This increased resistance at extremes of piston movement is very important and coincides with toe off, preventing excessive heel rise, and late swing preventing terminal impact.

The Dynaplex 'H' knee unit utilizes two chambers and operates via a piston forcing a fluid through a single adjustable orifice into a second chamber which incorporates a spiral spring. The characteristics of fluid flow turbulence is coupled with the characteristic of a spiral spring, which offers greatest resistance to compression at initial compression and near total compression. Upon flexion and initial piston movement the fluid is forced into the second chamber where it acts on the spring and is also forced to the opposite side of the piston, again, in preparation for the pistons return. Upon the pistons return, the spring forces the fluid into the first chamber and hydraulic damping provides resistance to full extension and the pistons extremes of movement. This, once again, is important as this unit offers greatest resistance at toe off and late swing.

The mechanics of 'H' units results in a knee mechanism which when correctly set at a comfortable walking speed allows less knee flexion at toe off and automatic responsiveness to fast and slow walking.

A knee which is automatically cadence responsive.

A 'CT' knee controls swing phase via a mechanical means, that is, using mechanical friction. This mechanical friction acts as a means of simply slowing the pendular swing of the shank. A pendular swing time which is a result of the mass of the shank and where in the shank the Centre of Gravity is located. The level of friction set into the knee unit is determined by the patients comfortable walking speed, and essentially is set to minimize terminal impact at the end of swing. This friction setting for comfortable walking only, therefore restricts the walking speeds available to the amputee.

Definitely not a cadence responsive knee mechanism.
The behaviour of the prosthetic shank in an Above Knee prosthesis using a "CF" knee is particularly likened to that of a simple pendulum. Skinner (1980) and Tashman et al. (1987) have proven that by shifting the Centre of Gravity (which is also the Centre of Rotation) nearer the knee centre, a decrease in the swing period was noted. Just as moving the weight higher on the pendulum of a Grandfather clock leaves the clock ticking that little bit faster.

The simple pendulum as defined by an array of texts has a period of oscillation which is dependant on the Radius of Gyration.

\[ T = 2\pi\sqrt{\frac{R}{G}} \]

where, \( T \) = Period of Oscillation
\( R \) = Radius of Gyration
\( G \) = acceleration via gravity

Equation (1)

For the purposes of this research and for swing phase duplication the "H" knee swing phase will also be considered a simple pendulum, although it is recognised that it is not. The "H" knee has phases of high friction at both the beginning and end of swing, limiting knee flexion at toe off and terminal impact before heel contact. Calculations of the Radius of Gyration for the "H" knee were made through finding the period of oscillation for the "H" knee at three walking speeds. The calculated Radius of Gyration was duplicated when using the "CF" knee and achieved via the placement of a mass.

This fact lead one to presume that the shifting of the Centre of Rotation proximal and distal from a set position for comfortable walking would result in a shank section that essentially had three gears. Upon wanting fast walking, the amputee would simply shift a moveable mass to a precalculated proximal location and likewise distal for slow walking.

A manually operated cadence responsive "CF" knee giving the same results as a "H" knee but at a lower cost.
Equipment

A prosthetic leg was manufactured via duplicating the subject's current socket. The socket was mounted on a USMC Ultralite Multiplex Mark V Modular System with a standard SACH foot.

A USMC Dynaplex Hydraulic Gait Control Unit was used as the Hydraulic Knee Unit.

A USMC Mechanical Extension Bias and Resistance Control Unit was used as the Constant Friction Knee Unit.

A High Speed Cinematographic Camera was used to record the gait of the subject.

A Stop Frame Projector and Film Digitizer were used to analyze the gait of the subject.

A set of accurate scales was used to ascertain knee unit mass and the mass of additional weight.

Douglas bags were used as a means of recording oxygen uptake per minute; this was used to evaluate the energy efficiency of the different shanks.

A stopwatch was used to time walking.
Procedure

Values for the resting Centre of Gravity (Radius of Gyration) for the shank incorporating the "H" knee unit and then the "CF" knee unit were taken. These figures taken as being the actual resting Centre of Gravity's.

Cinematographic film was made of fast, free-speed and slow walking with the prosthesis incorporating the Hydraulic Knee unit. Upon analysis of the film via the stop frame projector and digitizer, figures for Velocity, Cadence, Stride length, Knee Angle, Stride time and Swing phase were calculated. Treating the Hydraulic shank as a simple pendulum the Swing time could be doubled to arrive at a figure for the period of oscillation of the shank. This was carried out for the three different walking speeds.

Each period was substituted into equation (1) to arrive at figures giving the Radius of Gyration of the shank at the three different speeds.

The "CF" knee unit was modified so as to delete the extension bias, this was achieved through removing the extension spring. This gave a "CF" swing phase dependant on pendular activities.

Cinematographic film was made of fast, free-speed and slow walking with the prosthesis incorporating the "CF" knee unit. Data collected from the analysis of this film was used to prove the differences between gait speeds and symmetry of the "H" and "CF" knee units suggested by Murray (1983).

The three values for Radius of Gyration for the "H" knee were attempted to be duplicated with a mass equal to the mass difference between the two knees (175 gms) and it was found to be impossible. With this weight it was possible to duplicate the actual resting C.G of the "H" shank and this was later used as that for slow walking.

A mass of 750 gms was used to duplicate the values for the Radius of Gyration for the "H" knee at fast and free speed walking, however duplication of slow walking was not possible as this required the mass to be placed on the heel of the shoe, a position not satisfactorily achieved and hence abandoned. Using this greater weight, duplication of the actual resting C.G of the "H" shank was also achieved, and due to excellent results also used as that for slow walking.

Cinematographic film was made of walking with the constant friction knee unit with a mass positioned appropriately to gain a centre of gyration equal to that of the Hydraulic shank at the three different walking speeds. Analysis of this film gave figures for Velocity, Cadence, Stride Length, Knee angle, Stride Time and Swing Phase. These figures were compared to those of "H" walking to ascertain whether successful gait duplication had been achieved.

Oxygen consumption was measured for walking with the "H" knee and with the modified "CF" shank to distinguish whether the adapted shank required more energy input than did the "H" unit in giving the same gait.
Results

- **Calg Duplication**
  - Through the values of swing time for the "H" knee walked at the three speeds, and using equation 1, the following values for the Radius of Gyration for the "H" knee were found.

<table>
<thead>
<tr>
<th>Table 1. R.O.G. for &quot;H&quot; knee at differing speeds</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow Walking</td>
</tr>
<tr>
<td>---------------</td>
</tr>
<tr>
<td>R.O.G (1 mm distal to knee center)</td>
</tr>
</tbody>
</table>

These values were duplicated with the "CF" knee through the addition of the lead mass (170 g) in several positions. The duplication of slow walking with the "CF" knee was in fact a duplication of the "H" knee's actual R.O.G. this duplication was possible with both the lead mass and the plastacene mass (179 g) which represents the mass difference of the two knee units.

- **Table 2. Mass position for duplication of "H" R.O.G.**

<table>
<thead>
<tr>
<th>Slow Walking*</th>
<th>Free Speed</th>
<th>Fast Walking</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lead Mass Position</td>
<td>270</td>
<td>216</td>
</tr>
<tr>
<td>Plastacene mass position</td>
<td>100</td>
<td>Duplication not possible</td>
</tr>
<tr>
<td>(1 mm distal to knee center)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* Duplication of actual "H" R.O.G.

Stride parameters recorded and calculated for the three different mechanisms trialed are represented in Table 3.
<table>
<thead>
<tr>
<th>Gait Measurement</th>
<th>Slow &amp;</th>
<th>Free Speed</th>
<th>Fast</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Velocity (cm/s)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant Friction</td>
<td>88</td>
<td>92</td>
<td>97</td>
</tr>
<tr>
<td>Hydraulic</td>
<td>96</td>
<td>106</td>
<td>110</td>
</tr>
<tr>
<td>Adapted Shank</td>
<td>101</td>
<td>110</td>
<td>113</td>
</tr>
<tr>
<td>Controls *</td>
<td>58</td>
<td>151</td>
<td>218</td>
</tr>
<tr>
<td><strong>Cadence (steps/min)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant Friction</td>
<td>82</td>
<td>82</td>
<td>88</td>
</tr>
<tr>
<td>Hydraulic</td>
<td>77</td>
<td>86</td>
<td>90</td>
</tr>
<tr>
<td>Adapted Shank</td>
<td>88</td>
<td>94</td>
<td>97</td>
</tr>
<tr>
<td>Controls *</td>
<td>51</td>
<td>113</td>
<td>138</td>
</tr>
<tr>
<td><strong>Stride Length (cm)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant Friction</td>
<td>128</td>
<td>132</td>
<td>132</td>
</tr>
<tr>
<td>Hydraulic</td>
<td>133</td>
<td>145</td>
<td>148</td>
</tr>
<tr>
<td>Adapted Shank</td>
<td>138</td>
<td>141</td>
<td>140</td>
</tr>
<tr>
<td>Controls *</td>
<td>126</td>
<td>156</td>
<td>186</td>
</tr>
<tr>
<td><strong>Cycle Duration (sec)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant Friction</td>
<td>1.46</td>
<td>1.46</td>
<td>1.36</td>
</tr>
<tr>
<td>Hydraulic</td>
<td>1.53</td>
<td>1.40</td>
<td>1.34</td>
</tr>
<tr>
<td>Adapted Shank</td>
<td>1.36</td>
<td>1.28</td>
<td>1.24</td>
</tr>
<tr>
<td>Controls *</td>
<td>1.48</td>
<td>1.06</td>
<td>0.57</td>
</tr>
<tr>
<td><strong>Swing Phase (sec) (3 Cycle)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant Friction - Sound</td>
<td>.52 (.35%)</td>
<td>.50 (.35%)</td>
<td>.42 (.30%)</td>
</tr>
<tr>
<td>- Prosthetic</td>
<td>.55 (.40%)</td>
<td>.66 (.45%)</td>
<td>.56 (.41%)</td>
</tr>
<tr>
<td>Hydraulic         - Sound</td>
<td>.53 (.36%)</td>
<td>.53 (.37%)</td>
<td>.49 (.36%)</td>
</tr>
<tr>
<td>- Prosthetic</td>
<td>.63 (.41%)</td>
<td>.56 (.40%)</td>
<td>.54 (.40%)</td>
</tr>
<tr>
<td>Adapted Shank     - Sound</td>
<td>.48 (.35%)</td>
<td>.50 (.38%)</td>
<td>.46 (.36%)</td>
</tr>
<tr>
<td>- Prosthetic</td>
<td>.56 (.41%)</td>
<td>.52 (.41%)</td>
<td>.54 (.41%)</td>
</tr>
<tr>
<td>Controls *        - Sound</td>
<td>.52 (.35%)</td>
<td>.41 (.39%)</td>
<td>.38 (.44%)</td>
</tr>
<tr>
<td><strong>Stance Phase (sec) (3 Cycle)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant Friction - Sound</td>
<td>.95 (.65%)</td>
<td>.94 (.65%)</td>
<td>.95 (.70%)</td>
</tr>
<tr>
<td>- Prosthetic</td>
<td>.88 (.60%)</td>
<td>.80 (.55%)</td>
<td>.80 (.59%)</td>
</tr>
<tr>
<td>Hydraulic         - Sound</td>
<td>.93 (.64%)</td>
<td>.59 (.63%)</td>
<td>.89 (.64%)</td>
</tr>
<tr>
<td>- Prosthetic</td>
<td>.92 (.59%)</td>
<td>.56 (.60%)</td>
<td>.80 (.60%)</td>
</tr>
<tr>
<td>Adapted Shank     - Sound</td>
<td>.88 (.65%)</td>
<td>.52 (.62%)</td>
<td>.82 (.64%)</td>
</tr>
<tr>
<td>- Prosthetic</td>
<td>.80 (.59%)</td>
<td>.76 (.59%)</td>
<td>.70 (.56%)</td>
</tr>
<tr>
<td>Controls *        - Sound</td>
<td>.96 (.65%)</td>
<td>.65 (61%)</td>
<td>.49 (56%)</td>
</tr>
</tbody>
</table>

* Values for Controls taken from Murray (1953)

Adapted shank values for slow walking taken with c.o.g. equal to resting Hydraulic shank c.o.g.

The behaviour of the adapted shank in comparison to the "CF" shank is best summarized as follows.

**Table 4 Summary of Adapted Shank Characteristics compared to the "CF" Shank**

<table>
<thead>
<tr>
<th></th>
<th>Free Speed</th>
<th>Fast Walking</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Velocity</strong></td>
<td>20%</td>
<td>17%</td>
</tr>
<tr>
<td><strong>Cadence</strong></td>
<td>15%</td>
<td>10%</td>
</tr>
<tr>
<td><strong>Prosthetic Swing Time</strong></td>
<td>21%</td>
<td>4%</td>
</tr>
</tbody>
</table>
Graphs one and two demonstrate the behaviour of the adapted shank in comparison to both the "H" knee and the "CF" shank in terms of Velocity and Cadence, two parameters which clearly demonstrate the improvement in "CF" gait achieved with the addition of mass to the shank. Control values taken for normal from Murray (1983).

**Graph 1. Walking Velocity V's Walking Speed**

**Graph 2. Cadence V's Walking Speed**
Ideally the gait of an amputee should be as symmetrical as the non-amputee however this is rarely the case. Murray (1983) found that “H” knee mechanisms gave a gait far more symmetrical than a “CF” knee. As is the case with this program, the “H” unit gave a more symmetrical gait than the “CF” knee and surprisingly the adapted shank gave a more symmetrical gait than did the “H” unit. (Figure 1) Results beyond expectations and arising great scepticism. However, it was found conclusively that at fast walking the adapted shank gave a faster stride time than the “H” unit and a percentage break down of prosthetic stance and swing equal to the non-amputee. Doubt does arise when viewing the same figures for the “H” and “CF” units giving percentile values of swing less than the non-amputee results indicating the need for more trials and comprehensive results from which means could be taken.

<table>
<thead>
<tr>
<th>Normal Men</th>
</tr>
</thead>
</table>
| Stance: 56% | Swing: 44%  
| LHC | Stance: 56%  
| RHC | Swing: 44%  
| LHC | Stance: 56%  
| RHC | Swing: 44%  

<table>
<thead>
<tr>
<th>Female Ak Amputee with Adapted Shank</th>
</tr>
</thead>
</table>
| Stance: 56% | Swing: 44%  
| PHC | Stance: 64%  
| SHC | Swing: 36.5%  
| PHC | Stance: 64%  
| PHC | Swing: 36.5%  

<table>
<thead>
<tr>
<th>Female Ak Amputee with Hydraulic Knee Unit</th>
</tr>
</thead>
</table>
| Stance: 60% | Swing: 40%  
| PHC | Stance: 64%  
| SHC | Swing: 36.5%  
| PHC | Stance: 64%  
| PHC | Swing: 36.5%  

<table>
<thead>
<tr>
<th>Female Ak Amputee with Const. Friction Knee Unit</th>
</tr>
</thead>
</table>
| Stance: 39% | Swing: 41%  
| PHC | Stance: 70%  
| SHC | Swing: 30%  
| PHC | Stance: 70%  
| PHC | Swing: 30%  

![Figure 1](的关键部分) Presentation of Gait Symmetry for the different shank mechanisms.
taken from taken from Frankel (1984), it was found that the adapted shank demonstrated a pattern similar to the "H" shank in most cases and in the case of free speed walking a considerable decrease in knee flexion as compared to the "CF" shank. These results are shown in graphs 4, 5, 6.

Graph 4. Knee Angle vs. % of Cycle for Slow Walking

Graph 5. Knee Angle vs. % of Cycle for Free Speed Walking

Graph 6. Knee Angle vs. % of Cycle for Fast Walking
Oxygen consumption was measured initially in a resting state and then measured upon completion of 3 minutes warming up and 2 minutes of free speed walking with the IF unit.

Using the adapted shank with the additional mass set for free speed duplication, walking was to be carried out for another two minutes. However, after warming up for 3 minutes only 1 15 minutes of walking was possible due to the "CF" knee seizing.

Over this period of free speed walking with the "CF" knee the heat generated via the friction mechanism had expanded the materials sufficiently to prevent knee motion. A problem which resulted in an awkward gait and a gait one would expect to require excessive energy.

Graph 3 represents the Oxygen uptake per kilogram per minute for comfortable walking and such figures compare favourably to results from Waters et al (1976). His work was with 70 patients and found that the traumatic amputee (usually quite fit and healthy) used as a comparison to the young, fit and healthy neoplastic amputee used in this program had an average Oxygen intake for walking of 12.0 ± 3.4 ml/kg/min, a figure which both the measured values fell well within. The shaded region represents one standard deviation for the results found by Waters et al (1976).

**Figure 1. Oxygen Consumption for Free Speed Walking**
Discussion

Gait duplication with the adapted shank was extremely successful and gave values of Velocity, Cadence, Stride Length, Cycle Duration, Knee angles, Swing time, Stance time, and Oxygen Consumption near the \( \text{H} \) knee and better in many cases.

Duplication not being possible with the plastacene mass that is, with both shank sections being of identical mass, demonstrated that the mass and position of mass of the \( \text{H} \) unit in the shank has little to no bearing on the gait pattern achieved. This was very interesting and immediately ruled out the possibility that the \( \text{H} \) mechanism operates second to the effect the mass of the unit has on the shanks pendular characteristics.

Successful gait duplication with the greater mass of \( 790 \) grams, highlighted the and importance of the centre of mass of the prosthetic shank and how for many years the prosthetist may have inadvertently hindered an amputee's optimum velocity by unknowingly manufacturing a shank with an excessively distal centre of mass. Presently a gradual awareness of shank mass and mass location is becoming evident in the profession and I believe will continue.

Results for the Adapted shank, as for the "H" shank gave results which supported the work of Murray (1983), the adapted shank achieved a greater walking velocity through an increase in cadence rather than an increase in stride length.

Cycle duration for the adapted shank was nearer the control values than the "H" or \( \text{CF} \) shanks for free speed and fast walking, giving an increased gait symmetry. Values for swing phase and stance phase were that of the control values for the prosthetic leg and as expected a little more sound stance to compensate for prosthetic insecurity. An interesting matter which can be attributed to errors or lack of repeated testing was the prosthetic swing phase percentage achieved by both the "H" and "CF" shanks. Results indicating that the prosthetic shank was swinging faster as a percentage of stride time than the non amputee. Results which are exceptionally excellent and thus worthy of some questioning.

Examination of the knee angle graphs, demonstrated, as reported by Murray (1983) that \( \text{CF} \) knees tend to have an increased knee flexion angle at toe off, compared to "H" knees, thus slowing the swing time and walking velocity. This is best demonstrated at slow and fast speed walking. It is surprisingly found that at fast walking, all shank mechanisms follow essentially the same knee flexion pattern, a walking speed which one would predict to show the greatest variance of results.

Oxygen consumption (a measure of energy expenditure) for the "H" and the adapted shanks proved very similar and what variance there was in these results is rather negligible and attributed to the awkward gait whilst recording data due to the seizing of the "CF" knee.
Conclusion

In summary this research has laid the groundwork for much further study, a field of study which through this program and others have proven that the gait pattern of an Above Knee amputee is essentially predicted by the pendular characteristics of the prosthetic shank. Previous work had shown the importance of the pendulum in the swing phase of the prosthesis, yet no work had been directed at the possibility of achieving the gait pattern of an expensive and complex unit with a not so expensive and simple unit. This project found answers to many questions and even answers to questions not thought of.

Walking velocity can be increased or decreased through manipulating the shank sections centre of mass, but not only changed, it can be predicted with quite a degree of accuracy and set to a specific desired velocity. Velocity increases and decreases are not a result of a change in stride length, but because of a change in cadence, that is, the ability to take more or less steps per minute.

Symmetry of walking was increased with faster walking. Not only was the gait pattern of the H knee duplicated but a more symmetrical gait was achieved, a gait which was exceptionally close to the non amputee male.

All these findings were very rewarding yet would have proven pointless if it required excessive amounts of energy to achieve "H" gait with a "CF" knee. The degree of oxygen consumption for the adapted shank proved that the project was viable and extremely worthwhile.

Closing it can be said that this project has opened the door to, hopefully, much more research into this area and with the rising prices of prostheses and componentry it would appear that soon very few will be able to afford the best knee mechanisms and an alternative will be sought. When this day comes, and I believe it will, there is no need to cry out for alternatives.

An alternative already exists.

Acknowledgements

I would like to thank David Orr and Les Barnes for their help with this project and the tireless efforts of Louella Sharrock, without whom this research would not have been possible.
Bibliography

1. APYA, Atam P.
   "Introductory College Physics".

2. BENNETT WILSON A.
   "Hydraulics and above knee prosthetics".
   Clinical P&O Vol 7 No 4 Fall 83, 3-4.

3. FRANTZ J.
   "Basic Biomechanics of the Skeletal System".
   1980.

4. GODFREY C M, JOUSSE A T, BRETT R, BUTLER J F.
   "A comparison of some gait characteristics with six knee joints".

5. GODFREY C M, JOUSSE A T, BRETT R.
   "Foot mass effect on gait in the prosthetic limb".

6. INMAN V T.
   "Conservation of energy in ambulation".

7. ISHAL G, BAR A.
   Evaluation of AF prostheses comparing conventional with adaptive knee control devices.

8. FREIDRICH D, TASHMAN S.
   "Kinematic and kinetic comparison of the conventional and ISNY AK socket".
   Clinical P&O Vol 9 No. 3, 28-36.

9. LANHAMMER H.
   "Variation of mechanical energy levels for normal and prosthetic gait".

10. LEACH P G L.
    "The pendulum in classical and Quantum Mechanics".
    Templestowe, 1975.

11. LEHNEIS H R, SUNG CHU D., ADELGASS H.
    "Flexible prosthesis socket techniques".
    Clinical P&O Vol 8, No. 1, pp6-11.

12. LEWIS E A, BERNSTOCK, W M.
    "Clinical application study of the Henschke-Mauch Hydraulik Mode A Swing and stance control system".
    Prosth Res Fall 80, pp33-38.

13. MUILENBERG A L.
    "Basic changes in lower limb prosthetics".
    Clinical P&O Vol 9 No 4.
MURPHY E F
The swing phase of walking with above knee prostheses

MURPHY E F, HOLLON L, ZITZEWITZ D, SMOOT C
Physics principles and problems
Charles E Merrill publishing co. 1972.

MURRAY M P, MOLLINGER L, SEPICS S, GARDNER G, LINDE M
Gait patterns in AK amputee patients: Hydraulic swing control versus constant friction knee components
Arch Phys Med Rehabil Vol 64. Aug '83.

MURRAY M P, MOLLINGER L, SEPICS S, GARDNER G, LINDE M
Gait patterns of AK amputees using constant friction knee components

NOLLEN AND PARKER J
Advanced level physics

RICHARDS L, SEARS W, WEHR A, ZEMANSKY M
Modern University Physics

SADOLICH J
The 0-1-2-3 Above knee running system

SKINNER H B, MOTIE Jnr C D
Optimization of amputee prosthesis weight and weight distribution
Rehabilitation R&D Progress Reports. 1984.

STAROS A. MURPHY E F
Properties of fluid flow applied to above knee prostheses

TASHMAN S, HICKS R, JENDRZEJECZK D J
Evaluation of a Prosthetic Shank with Variable Inertial Properties

VAN DE VEFN P G, VAN DER TEMPEL W, DE VREISS J
Bond graph modelling & simulation of the dynamic behaviour of above knee prostheses
Prosthetics and Orthotics International 1987. 11. 65-70.

WATERS R L, PERRY J, ANTONELLI D, HISLOP H
Energy cost of walking of amputees: The influence of level of amputation