APPLICATION OF MECHANICAL MODELING TO SPORT BIOMECHANICS

Richard Smith
The University of Sydney, Sydney, NSW, Australia

Mechanical modelling and particularly inverse dynamics methods can be useful in sports biomechanics. This aim of this paper was to illustrate how various models applied to rowing and running can improve understanding of the movement, performance and reduce injuries. Simple and complex models were used. Measures of power were effective in elucidating important mechanisms in the rowing action and in identifying areas of improvement in rowing technique. Differences between ergometer types in the simulation of rowing were clear from inverse dynamics analysis and the addition of a muscle model for the lumbar spine provided useful suggestions for the reduction of lumbar stress. The application of the same analysis techniques to other situations was illustrated by a running and footwear example.

KEY WORDS: mechanics, modeling, rowing, footwear, performance, injury.

INTRODUCTION: The application of biomechanical principles to sport can improve the understanding of movement mechanisms, assess and improve performance, and provide a knowledge base for schemes to prevent injuries. The aim of this paper was to provide some examples of the application of mechanical modeling to sports biomechanics in these three areas and to discuss their limitations and how these limitations might be overcome.

The simplest models consider the human body to be concentrated at a point. This could be usefully applied in race analysis. A more complex model is the linked segments model which may contain from one to 17 or more segments linked by joints. The models may be two- or three-dimensional and the joints may have from one to six degrees of freedom. Usually the segments are assumed to be rigid, connected by ideal joints, with masses and moments of inertia that do not change during movement Linked segment models may be used to deduce joint torques and forces from kinematic information (inverse dynamics) or the kinematics may be deduced from the application of joint torques to the model (forward dynamics). In the latter case the torques may be derived from mathematical models of the muscles acting on moment arms at the joint. In choosing a model one should use the principle attributed to Albert Einstein: "Make the model as simple as possible but not simpler." Only inverse dynamics models will be considered here. Joint power can be integrated over time to obtain the work done at the joint. In doing so it is assumed that no energy is required to overcome antagonistic activity of the muscles, move the muscles themselves relative to the skeleton, work against elastic and non-elastic internal forces, or to maintain isometric force. Notwithstanding these assumptions, it can provide information about the fraction of muscle power expended to overcome external resistance and to change the mechanical energy of body segments. Simple and complex models of rowing will be used to illustrate the use of these mechanical models.

METHODS: For the on-water study, three-dimensional pin and stretcher forces, boat velocity and acceleration and seat position were measured during on-water rowing (Smith and Loschner, 2002). Handle force, stretcher forces, and electromyograms were recorded during ergometer rowing. Body segment positions were recorded in two dimensions during on-water rowing and in three dimensions during ergometer rowing. Two dimensional, 9-segment, inverse dynamics analysis was carried out using custom software. The ergometer rowing analysis was extended to a muscle and joint model for the lumbar region of the spine to estimate lumbar compressive force. For the running study rearfoot motion was measured barefoot and inside shoes using a wand marker system mounted on the calcaneus (Kinchington and Smith, 1996) and skin-mounted markers on the shank, thigh and pelvis. An eight camera motion analysis system and force platform was used to collect position and force data. The data were used in an inverse dynamics program to determine the important kinematic and kinetic variables.

RESULTS AND DISCUSSION: An application of the point mass model to performance during the recovery phase of on-water single sculling will be considered first. Two biomechanical principles are applied. Firstly, the fluctuations in boat velocity should be kept small to minimise energy losses against drag. Secondly, a force is required to accelerate a body. During the recovery phase foot contact with the stretcher is the only means of applying a propulsive force to the boat. This can be achieved by accelerating the rower's body towards the stern of the boat creating a reaction force towards the bow.

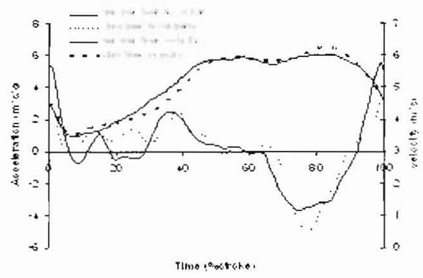

Figure 1: Seat acceleration and boat velocity (n = one national and one club level rower).

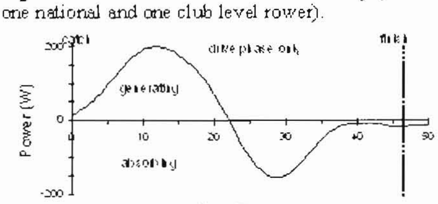

Figure 2: Power output at the knee joint (mean of n = 10 female ergometer rowers).

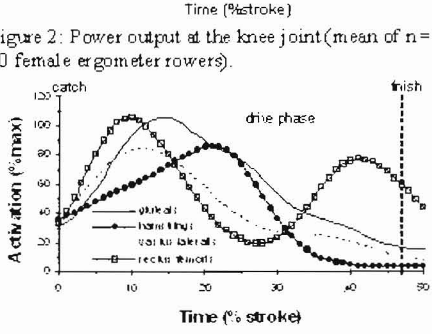

Figure 3: Activity of the lower limb muscles (mean of n = 10 female ergometer rowers).

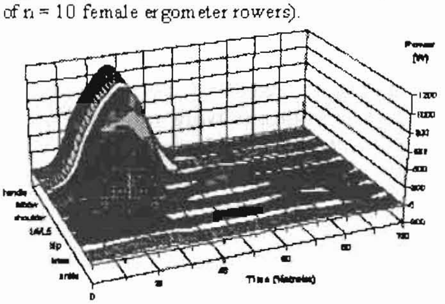

Figure 4: Cumulative power crossing joints (mean of n = 10 female ergometer rowers).

In this case the rower was considered as a point mass located at the seat. Applying the point mass model, the seat acceleration should be constant during the recovery phase to maintain a constant net force on the boat (Figure 1). This is impossible, but a skilful rower can approach the ideal. The national level rower was able to maintain a more constant acceleration from about 63 to 92% of the stroke compared with 69 to 88% for the club level rower. The resulting boat velocity was more constant during this period for the national level rower. The simple model did not take into account the trunk movement which can occur independently of the seat. Thus the model should be made more complex.

Using inverse dynamics calculation of joint power to understand lower limb function during maximal ergometer rowing is the second example. Power developed at the knee joint (Figure 2) showed a large region of power absorption between 22 and 35% of stroke time. Considered in conjunction with the electromyograms (Figure 3) there was strong evidence for transfer of this power back to the hip via the hamstring muscles. The amount of energy transferred was about 60 J.

The third example is from the same study and compares the timing and magnitude of power production at the major joints. Net power developed at each joint was eventually transferred to the handle (the ergometer stretcher was stationary). The magnitude of the power varied in magnitude and time (Figure 4) with the hip extensors producing the most power followed by the knee, shoulder (rotation), L4/L5, ankle, shoulder (translation), and elbow. The effect on handle force of transfer to the kinetic energy of the trunk can be seen. At the beginning of the drive phase (0 %stroke), although power was being developed in the lower limb it was not appearing at the handle. Part of the energy stored in the trunk was passed on to the handle by the arms after 40 %stroke.

Example four is the application of inverse dynamics to the calculation of mechanical energy expenditure due to the joint moments (MMEE). The MMEE for each joint can be obtained by integrating the power for that joint with respect ot time. The MMEE can be summed over all joints to arrive at the total MMEE for the rower. The group of 10 female rowers consisted of five national level and five club level rowers. They rowed maximally for six minutes and the total MMEE for each rower was measured during each minute. Differences between levels of rowers were evident (Figure 5). A significantly greater amount of MMEE was required by the club level rowers to produce each joule of external work than by the national level rowers (p = 0.038). Furthermore, the amount of MMEE required to produce each joule of external work increased with time (p = 0.003). The reason for the difference and trend over time was evident on examination of the differences at the joint level. The club level rowers had a greater degree of energy absorption at the joints than the national level rowers and both groups increased the amount of energy absorbing events with time and fatigue.

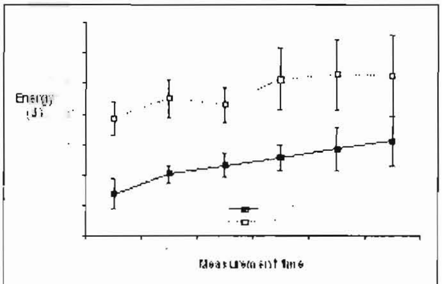

Figure 5: Means and standard error bars for the moment mechanical energy expenditure per joule of external work of the national and club level rowers.

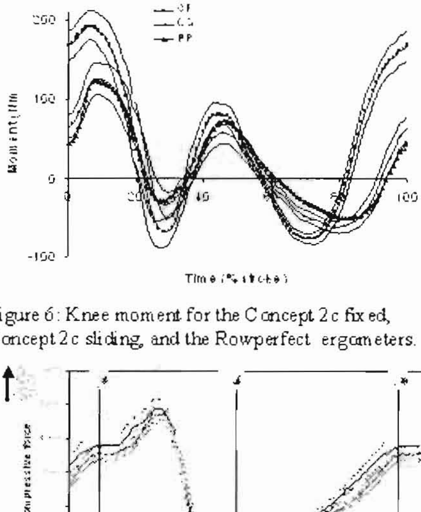

Figure 6: Knee moment for the Concept 2c fixed, Concept 2c sliding, and the Rowperfect ergometers.

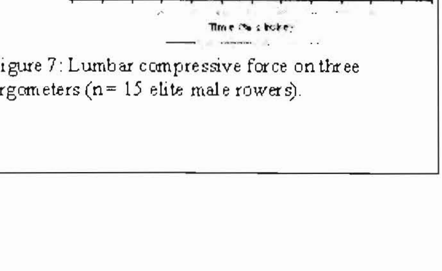

Figure 7: Lumbar compressive force on three ergometers (n= 15 elite male rowers).

The fifth example is to do with simulation of on-water rowing. There are three ergometer types in common use around the world. The Concept 2C fixed (stationary ergometer), the Concept 2C on slides (moving ergometer) and the Rowperfect (moving fan/stretcher assembly). On all ergometers the maximum acceleration occurred at the catch and finish. On the fixed ergometer this acceleration is shared between the mass of the earth and the rower's body. On the moving ergometers the acceleration is shared between the moving part of the ergometer and the rower's body. The moving part of the ergometers is lower in mass than the rower. This creates a difference in the stretcher reaction forces and causes differences in the joint moments, especially of the knee joint (Figure 6). The curves are shown with 95% confidence intervals and it was clear that the Concept 2C fixed ergometer was associated with twice the knee joint moment near the catch (0-10 %stroke) and at the end of the recovery phase.

The sixth example follows logically from the fifth with the question: What effect does the ergometer type have on lumbar compressive force? Using a lumbar extensor muscle model of the spine combined with knowledge of the lumbar joint moment the compressive force was estimated for the period when the spine was experiencing an extension moment (Figure 7). The lumbar compressive force was significantly greater for the concept 2C fixed condition than the other two ergometers for the first and last 15% of the whole stroke. Rowing for more than 30 min on an ergometer was identified as a consistent predictor of low back pain in intercollegiate rowers (Teitz et al., 2001).

The seventh example illustrates another application to prevention of injury but in this case to footwear. Bellchamber and van den Bogert (2000) in a barefoot running study queried the potential effectiveness of orthotics in shoes for those wearers who controlled their foot motion by proximal muscle action. We followed up this question by examining the power associated with tibial torsion during running at 3.8 m/s in three footwear conditions: barefoot, regular shoes, motion control shoes.

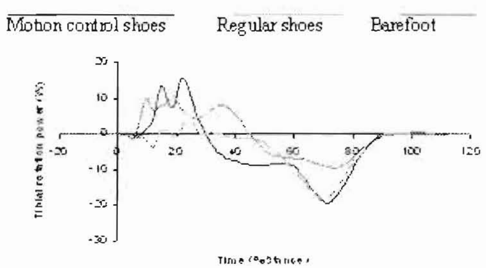

Figure 8: Power flow of tibial rotation in three footwear conditions (n = 10)

Results showed that motion control shoes reduced maximum eversion by 3.9o compared to barefoot running (P = 0.007) (Table 1). Also, external tibial rotation occurred much earlier in motion control shoes when compared to barefoot running (P = 0.019) and occurred while the knee was still flexing. Positive power flow up to 35% and 50% stance respectively for barefoot and motion control shoes (Figure 8) indicated that calcaneal eversion originated from the distal segment at early stance but the regular shoes had near zero power during this period of time. Thus the mechanics of the ground-shoe-rearfoot controlled the motion in the case of the barefoot and motion control shoe cases for most of the pronation period but was indeterminate for the regular shoes.

Table 1 Magnitude & timing of maximum eversion & tibial rotation under three footwear conditions.

Variable	Motion control			Regular			Barefoot		
Maximum eversion (degrees)	-7.0	±	1.4	-10.8	±	1.4	-10.9	±	1.3
Time to max eversion (%stance)	-46.6	±	2.7	47.3	±	2.9	47.1	±	2.7
Maximum internal tibial rotation (degrees)	-7.5	±	1.4	-10.7	±	2.3	-10.4	±	1.2
Time to max internal tibial rotation (%stance)	37.9	±	3.0	41.3	±	2.9	47.2	±	3.0

CONCLUSION: The research experience related above suggests that mechanical modelling and particularly inverse dynamics methods can be useful in sports biomechanics. The simple point mass model was sufficient to demonstrate efficient technique during the recovery in rowing and also showed the limitations of a simple model. Measures of power were effective in elucidating important mechanisms in the rowing action and in identifying areas of improvement in rowing technique. Differences between ergometer types in the simulation of rowing were clear from inverse dynamics analysis and the addition of a muscle model for the lumbar spine provided useful suggestions for the reduction of lumbar stress. The same analysis techniques can be applied to other situations as illustrated by the footwear example.

REFERENCES:

Bellchamber TL, van den Bogert AJ (2000). Contributions of proximal and distal moments to axial tibial rotation during walking and running. Journal of Biomechanics 33(11): 1397-1403.

Kinchington M, Smith RM (1996). An experimental design to enable the three-dimensional analysis of rearfoot mechanics whilst in footwear. The Australian Conference of Science and Medicine in Sport, Canberra, 28-31 October, 224-225.

Smith RM, Loschner C (2002). Biomechanics feedback for rowing. Journal of Sports Sciences. 20(10), 783-791.

Teitz, C. C., J. O'Kane, B. K. Lind, and J. A. Hannafin. (2001). Back pain in intercollegiate rowers. Am. J. Sports Med. 30:674-679.