EVOLUTION OF THE MEASUREMENT OF BODY SEGMENT INERTIAL PARAMETERS SINCE THE 1970S

Introduction

Since the development of biomechanics as a sub-discipline within movement science in the last 35 to 40 years (1), analysis techniques have evolved rapidly. To attain the goals of sports biomechanics - performance enhancement, comfort, injury prevention and safety (2) - it has been necessary to further develop techniques to both quantify and analyse data. Research questions have evolved from quantification of movement to questioning how and why movement occurs, and optimisation of performance. Methods of reconstruction such as the 3D Direct Linear Transformation (DLT) (3) and 2D-DLT (4) have evolved from creation to determination of the most accurate reconstruction method (5). Motion analysis has evolved from force-time data (6) to online systems and real-time feedback (7). Errors from soft tissue motion are now investigated to quantify and correct (8-9). Data smoothing has evolved from Winter et al.'s original paper on removal of kinematic noise (10) to modern work by Robertson and Dowling (11) investigating optimal filter design. Computer modelling has evolved from simplistic models of the 70s and 80s investigating simple locomotion (12) to sophisticated modern models of high bar gymnastics (13), high jump (14) and muscle stiffness of the horse (15). Initial work on co-ordination by Bernstein (16) has now evolved into a distinct field of motor control (17-18), with its own measurement issues (19). The focus of this article, however, is on the evolution of measurement techniques for determination of body segment inertial parameters (BSIP) with particular emphasis on development of mathematical models and scanning and imaging techniques.

How can inertial parameters be determined?

The three main methods of determining inertial parameters are cadaver studies (20-21), mathematical modelling (22-25) and scanning and imaging techniques (26-31). Each has its own advantages and disadvantages, and continuously evolved over the past 40 years. Early work by Dempster (21) calculated segmental masses as a percentage of total body mass, segment density, locations of the centre of gravity and lengths of radii of gyration as proportions of segment length based on eight elderly male cadavers and was built upon by Barter (32), Clauser et al. (33) and Chandler et al. (34). Chandler et al. (34) used palpable anatomical landmarks to identify the ends of segments (for example, the radiale was defined as the point at the proximal, lateral border of the head of the radius. This point was located by palpating downward in the lateral dimple at the elbow and getting the subject to pronate and supinate the forearm slowly so the radius could be felt rotating under the skin). This was in contrast to the subjective determination of joint centres as used by Dempster (21). Cadavers are normally not population-specific as they are typically of elderly males, quite different to any athletic or clinical population being studied. An increased demand for subject specific parameters hence led to development of mathematical models whereby the body is represented as a series of geometric solids; measurement of anthropometrics (segment width, depth or perimeter) allows for calculation of segment volume which when combined with density values (typically from cadaver studies such as Dempster (21)), permit calculation of segment mass.

Why are these parameters important?

Determination of BSIP such as mass, centre of mass and moment of inertia is important due to their use in kinetics calculations (35) with researchers wishing to make data as subject-specific as possible (36). In a clinical population, kinetics may be used to aid monitoring of joints post-trauma or post-operatively, or identify areas of particular stress. It may be used to identify a particular gait pattern that predisposes a patient to injury. Until the mid-90s, researchers concentrated on devising inertia models (21-23, 25-27) but more recent work has attempted to quantify the importance of accurate inertial parameters (37-38) and developed geometric-mathematical models of increasing sophistication such as that of Cheng et al. (30) and Gittoes and Kerwin (39) (40). Inertial parameters vary as a function of age (41), contracture of the muscle (42), limb morphology (43), body composition (39) and sporting background (44) but relative importance of these variances is still unknown. Some authors (38, 45-48) reported low importance of uncertainty in segment parameters, whilst others (37, 49-50) reported large variation in subsequent joint kinetics. Rao et al. (51) observed modelling the body as simple geometric shapes largely affected BSIP values calculated, particularly in segments such as the foot; Hanavan's (22) simplistic foot model was improved upon by Hatze (23), by remodelling it as 103 unequal trapezoidal plates, each with non-linear varying density.

The evolution of mathematical models

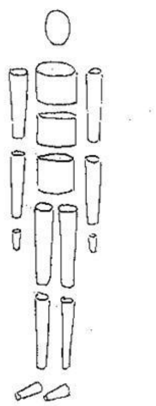

Figure 1. Hanavan's model (22). The simplistic modelling of each segment resulted in large error in calculation of mass

Hanavan's model (Figure 1) (22) was the first mathematical inertia model based on experimentally determined mass distributions and anthropometry of the person concerned, but had a number of limitations, namely the assumption all segments were rigid, of uniform density and uniform shape. High levels of inaccuracy were hence observed.

Hatze's model (Figure 2) (23) is presently the most accurate and reliable inertia model available (52). It used gender-dependent dependent density values, modelled the separate parts of the shoulder girdle, did not assume segment symmetry and realised the non-uniformity of segment shape. Hatze used the same segments as Hanavan (22) but with two additional shoulder girdle segments and alterations to hands and feet. The shoulder girdle, trunk segments and buttocks segments were geometrically quite complicated, with different gender-dependent density values used in the buttocks, thighs and calves. The 242 anthropometric measurements allowed for high levels of personalisation, but took 80 minutes to collect which obviously does not suit most subjects. No comparison of error in whole body moment of inertia (vertical axis) was made, where largest error may have been expected, with

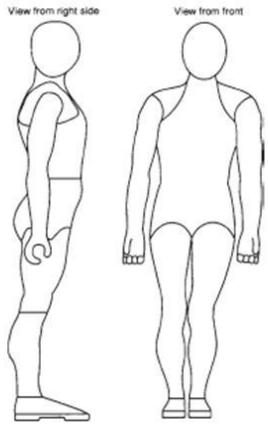

Figure 2. Segment definition as used by Hatze (23). The same segment definition was used for the left and right. A number of improvements to Hanavan's model (22) were made, amongst these being improved modelling of the hands as a prism and hollow half-cylinder to which an arched rectangular cuboid was added to represent the thumb

no evidence present that the author calculated moment of inertia about the anatomical vertical axis for the shoulder girdle. The method of removal of systematic error from the data was not reported. This model reports the lowest error between measured and predicted total body mass (0.5%), but its practicality is questionable.

The third, and perhaps most commonly used model due to its compromise number of measures, is that of Yeadon (Figure 3) (25), consisting of 40 sub segments and 95 anthropometric measurements, requiring 30-40 minutes contact time. Despite reported error of ~3%, (six times greater than Hatze (23)), Yeadon (25) considered the accuracy of his model to be sufficient due to reduced measuring time. The accuracy of the three models is comparable as they all used Dempster's density data (21) resulting in the only difference between them being volume measurement. The model was originally developed for use in gymnastics, assuming no movement at the neck, wrists or ankles hence limiting its applicability to sports with large motion at these joints. Yeadon represented the body as a series of stadium solids and truncated cones, with a semi-ellipsoid for the head, each of which represented their body part more accurately than that of Hanavan (22). The stadium represented trunk

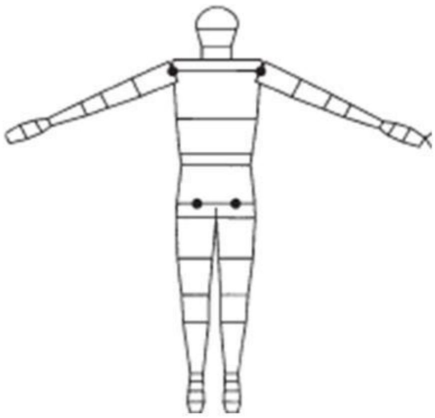

Figure 3. Yeadon's model, consisting of 40 sub-segments and requiring 95 anthropometric measurements (25)

volume particularly well (Figure 4), evolving from previous work modelling the trunk as an ellipse. A common limitation of all mathematical models, however, is that they cannot account for how soft tissue motion alters inertial characteristics of a segment (42) or the asymmetrical location of internal organs (53).

Figure 4. Cross-sections of the a) thorax b) stadium solid and c) ellipse. This clearly illustrates the improved representation of the thorax when modelled as a stadium as opposed to an ellipse (25)

The evolution of scanning and imaging techniques

Imaging techniques have been used to identify segment inertial parameters, with an acceptable level of success and removing potential harm from radiation exposure during gamma and DXA scanning. Jensen (26) digitised reference markers placed on segment boundaries and joint centres of three children of different somatotype (ectomorph, endomorph, mesomorph) and used Dempster's (21) data to calculate segment mass; this was the first paper to use a photogrammetric method. The body was modelled as a number of 0.2 cm high elliptical cylinders, the validity of which was confirmed by Wicke and Lopers (54). They also found increased image-size resulted in increased accuracy. This method was accurate to within 1.16-1.82% of total body mass, more accurate than other methods available at the time with further advantages that equipment was easily accessed, marker placement took ten minutes and digitising only two hours.

Later work by Hatze and Baca (55) and Sarfaty and Ladin (56) continued development of an image-based method of obtaining anthropometric dimensions. Baca (29) used a similar protocol to that of Jensen (26), with segment boundaries identified by use of black ribbon around the end which as noted by Sarfaty and Ladin (56) reduced magnitude of error. Use of sub-pixel accuracy reduced error from optical distortion, inaccurate edge-detection procedures and also user-specified upper and lower segment boundaries for edge-detection. Video images were gathered in the anterior, lateral (left and right side) and coronal views, and later processed by a programme called VIDANT which allowed for sub-pixel accuracy and edge estimation, initial estimation of segment edge, projection of distances in the reference plane, scaling of image dimensions to object space dimensions as well as correction of distortions. Segment dimensions obtained could be input into Hatze's 17-segment model (23) for segment volume determination and later combined with density data of Dempster (21) and Clauser et al. (33) to quantify segment mass. Maximum difference between the new image-based system and directly measured anthropometrics was 7.9% for the left forearm with average difference between 1 & 2%, implying the image-based method was a potential substitute for time–consuming direct measurement.

Gittoes et al. (31) further developed this work, digitising body segments based upon Yeadon's model (25). Peak Motus was used to digitise, which is more commonly used than the system of any other image-based study. Three photos (front of body, right and left lateral sides) were taken in a doorframe upon which 6 reflective markers were placed to allow for calibration. Even less contact time was required with the athlete as the segment boundaries were defined by eye, removing need for attachment of markers or ribbons. Digitising the points took approximately 30 minutes. Speed and decreased athlete contact time are the main advantages of this method. Difference between measures of total body mass as calculated by image and direct mass were found to be 2.10% and 2.87% respectively, supporting its suitability as an alternative to time-intensive direct measurement.

Scanning techniques were originally developed in the 80s and were seen as an evolution from Jensen's (26) and Hatze's (23) mathematical models. Zatsiorsky and Seluyanov (27) obtained mass and inertial characteristics from a gamma-scanning technique, but radiation exposure was an obvious disadvantage. Underlying tissue mass could be evaluated by intensity of absorption of the gamma-ray. The body was segmented into similar sized areas as previous cadaver studies to allow for comparison and segment mass, location of the centre of mass, radii of gyration, moments of inertia around the three axes and over 150 regression equations were determined. In contrast to cadaver studies, the shank and thigh were separated along the line of the knee joint (previous work often included a portion of the femur in the shank mass, distorting both shank and thigh mass), and the thigh was dissected from the trunk along a plane passing through the anterior superior iliac spine at an angle of 37° to the sagittal plane of the body. The equations developed are still seen today as the most accurate for kinetics calculations (35, 51).

More recent scanning studies have used computer tomography (CT) (57-59), dual X-ray absorptiometry (DXA) (60-61) or magnetic resonance imaging (MRI) (30, 62) due to minimal or, as with MRI, no, exposure to radiation. Pain and Challis (63) calculated BSIP calculation using a sonic digitizer, whilst Pinti et al. (64) determined BSIP using an optical scanner, with similar results to Jensen's model (26). Cost and limited access to equipment is a major disadvantage however. The validity of using DXA to obtain BSIP was determined by Durkin et al. (60) who calculated segment length, mass, centre of mass location and moment of inertia about the centre of mass for both a cadaver leg and a cylinder. Values obtained were cross-referenced to direct measurement to check accuracy and were found to be accurate and highly reliable. Two highlighted disadvantages were, however, radiation exposure and the two-dimensional nature of the image as only frontal plane data was available (the author acknowledged as data was gathered quickly, this was not an issue). Holmes et al. (61) further developed DXA, by determining fat mass, lean mass, wobbling mass and bone mineral content of the thigh, leg and leg and foot segments, applicable when modelling wobbling mass.

The evolution of wobbling mass models

Pain and Challis (63), during development of their high resolution method of calculating BSIP, found contraction of lower leg muscle altered BSIPs, particularly mass distributions. This change highlights a weakness of the rigid body model commonly used in biomechanical analysis, as redistribution of segmental mass may influence forces and moments at a joint. Both Gruber et al. (65) and Pain and Challis (42) found different forces were calculated using rigid body and wobbling mass models, with wobbling mass typically returning lower values. This highlighted a gap in the literature for a wobbling and rigid mass model to determine BSIPs.

Gittoes and Kerwin (39) hence designed one of the first models for determination of subject-specific BSIP for wobbling and rigid masses (Figure 5), which was applied to females and used in later work by Gittoes et al. (66). Using an adapted Yeadon model (25), 59 geometric solids (40 soft tissue, 17 bone, 2 lungs) were used to represent the components of the body to an accuracy of less than 3.0% (maximum error of 4.9%). The authors discussed the role lung volume and density variation during breathing played in model error, the limitations of the uniform density assumption (as investigated by Ackland et al. (58)) and the use of Dempster's density data. Ackland et al. (58) found intra-segment density data varied, suggesting future work may utilise methods such as DXA scanning to obtain personalised segment density values, which if used in conjunction with anthropometric dimensions may combine to produce a much more realistic and accurate model.

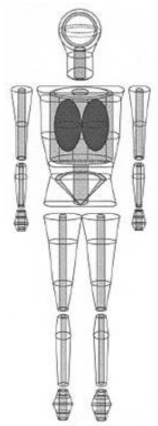

Figure 5. Segment and tissue distribution in the wobbling mass inertia model developed by Gittoes and Kerwin (39). Soft tissue is shown as white, bone as light grey and lungs as dark grey

What areas are currently being concentrating on?

Work this decade has developed simpler methods of BSIP calculation (68-70), and built on previous work (31, 71). Population-specific models are devised as researchers realise the generalised models of the 60s, 70s and 80s do not reflect the populations being investigated. Jensen was one of the first to recognise that the inertial properties of adults cannot be extrapolated to children; he published a number of papers (41, 72-76) investigating changes in children's inertial characteristics of children over a period of time. Jensen and Fletcher (77) and Pavol et al. (78) devised BSIP models of elderly adults, based on a growing trend of research into the biomechanics of older adults. Cheng et al. (33) determined BSIP of Chinese adults from MRI, whilst Nikolova and Toshev (79) calculated BSIP of the Bulgarian population using a 16-segment mathematical model.

Conclusions

Immense development of measurement of BSIP has occurred since the 1970s. Hatze's model (23) is currently the ideal, but its excessive time requirement limits its widespread use. Wobbling mass models are potential areas of growth in the future, and advances in measuring techniques such as DXA should aid this. Increasingly accurate and user-friendly techniques are likely to lead to more subject-specific models. Inertia modelling has been an important biomechanical analysis technique to date, and that importance looks set to continue into the future.

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