GEOFFREY DYSON MEMORIAL LECTURE

TRACKING VELOCITY IN MOTION

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TRACKING VELOCITY IN MOTION

It is a very great honor to be invited to present the Geoffrey Dyson lecture. It is many years ago now since I began studying his theories about human motion mechanics, and subsequently invited him to visit Wisconsin. At the time we were fortunate enough to have him come, he was theorizing about the possibility of angular momentum transferring from one body axis of rotation to another and was fascinated by our overhead view of a triple twisting. one and a half somersault dive. He wanted a copy of the film because it seemed to show rotation about a 3rd axis.

In choosing the title "tracking velocity in motion" I propose to trace my travels through a complex field, with. of course. the indispensable help of graduate students. I became interested in how the body generates and controls fast motions such as throwing, kicking, striking. etc., very shortly after I came to Wisconsin in 1948, following a stint in the British Army. It was Ruth B. Glassow who sparked my interest. As you may know, she was one of the first "Kinesiologists" who observed, photographed, analyzed and reasoned how it is we manage to perform such a range of difficult motor skills. It wasn't long before I was caught up in the search for cause and effect relationships, starting with basketball shooting. (Mortimer, 1950). Ruth was vitally interested in applying analytical results to teaching, thus, at that time, ideas went directly from the research laboratory to the teaching setting. We, as her graduate students, did the same, i.e., got a good idea from analyzing and went straight to the gym to teach it. I hope I didn't do too much damage, because as you will see. most of the "great" ideas I came up with turned out to be on the wrong track, or perhaps a side track. The title of my talk should really be "On the wrong track of velocity in motion".

One idea that was current when I was starting out was that a performer should try to make use of long body levers in generating high end point velocities. Thus, hip rotation and upper trunk rotation through their long lever arms were supposed to be important conmbutors to ball velocity in a throw and tennis forehand. Glassow thought that the time of release was an important time and analyzed lever contributions to ball velocity over the last ~ 25 msec. before release. Her analyses showed that pelvic rotation contributed 24% and the upper trunk 14% to release velocity in a skilled male performer: in a skilled female the values were 30% and 29% respectively (Cooper & Glassow, 1976). These measures taken from rather blurry, 64 fps film seemed to confirm the long lever principle and led to the conclusion that good technique involved keeping the hips and trunk rotating rapidly as the ball was released. i.e., performers should try to "follow through" with hip and trunk rotation.

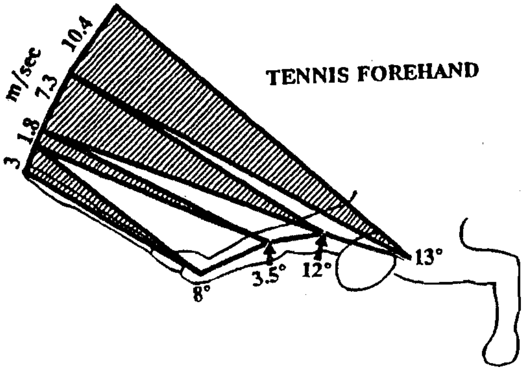


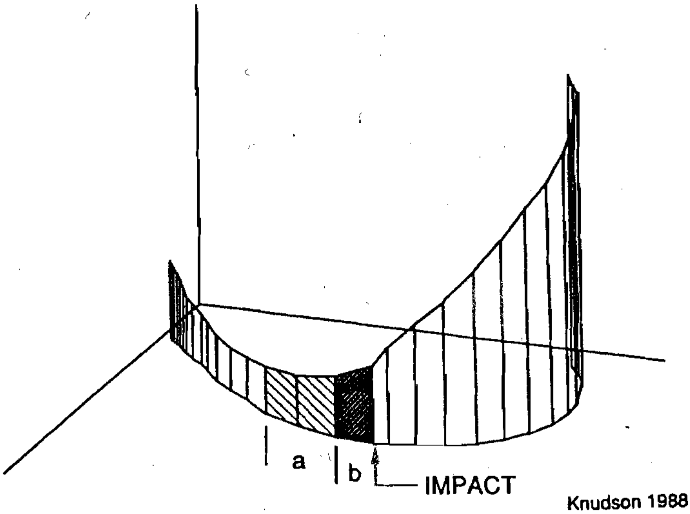
Figure 1. Top view of kinematic contributions of trunk rotation. horizontal adduction of humerus, elbow flexion and wrist action to linear displacement and calculated linear velocity of tip of tennis racket Thick black lines show moment arms and accompanying arc of motion for each angular motion. Longer moment arms provide greater linear velocities for the same degree of rotation.

Figure 1, an overhead view of a tennis forehand drive. illustrates how such measures were made. The moment arms for trunk rotation, shoulder horizontal adduction. elbow flexion and wrist action are shown together with degrees of rotation. arc of motion and the calculated linear velocity contributed by each action during a 30 msec interval near ball impact. The racket path shown in Figure 2. constructed from Knudson's (1988) recent data shows the approximate time period in question, (a). Trunk and hip rotation are combined in Fig. 1 with a mid-point axis, and showed a 13° rotation. Rotation of the humerus at the shoulder was similar at 12°.

However. the trunk with an ~ 30% longer moment arm contributes ~ 30% more to the arc of motion and so to the racket velocity. It can be seen that elbow flexion and wrist action were small contributors at this time. It wasn't until sometime later that I measured the kinematic contributions for the next time interval (b) i.e., over the 15 msec

TENNIS FOREHAND

PATH OF RACKET TIP



Knudson 1988

Figure 2. Three dimensional path of tennis racket tip in a tennis forehand drive: a) racket displacement over ~ 30 ms, b) displacement during –15 ms immediately before impact

immediately before impact.(*see* Figure 2). The contributions of each joint action proved to be quite different especially for the trunk and wrist. Trunk contribution decreased markedly to only 17% while wrist contribution increased to 35% from the previous ~ 15%.[1] The trunk rotation decrease is incompatible with the concept of trying to make use of its long moment arm. It would need to keep going to be most effective. Something must be wrong with the concept. As you probably already know, throwing has proved to have a similar problem. which I'll consider shortly.

In my own search for understanding I diverted for a time from movement kinematics to brain control mechanisms. Ruth Glassow's influence was partly responsible because she was interested in how the brain was controlling movement and had invited A.H. Steinhaus to give a workshop on the topic. In my teaching I had found that many students couldn't seem to follow the "marvelous" teaching cues I invented. Naturally, I couldn't think there was something wrong with the cues since they were based on "solid leverage principles"! Rather, I thought there must be something wrong with the students' brains; or perhaps that I wasn't tapping into their brain - movement connections. So I delved into brain research.

Shortly before that time (1955) the gamma motor neuron innervation of the muscle spindle had been discovered (Leksell, 1945). I thought it might tell us something exciting about muscle control. After five years of research, mostly under a high powered

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[1] **Shoulder rotation also increased to 41% from the previous 32%.**

microscope, we were able to demonstrate that the motor innervation of the muscle spindle had a representation in the motor cortex of cat and monkey much like that of skeletal muscle (Mortimer & Akert, 1961, Roberts, Smith, & Roberts, 1971). I was mighty excited about that thinking we now had an idea of how "kinesthesis" might control contraction. It was postulated that cortical messages might go to the spindle sense organ first (see Figure 3, II) then back through the spinal cord, via spindle afferents, and then to skeletal muscle (Merton, 1953). In this way other influences on spinal motor neurons could be integrated with cortical commands.

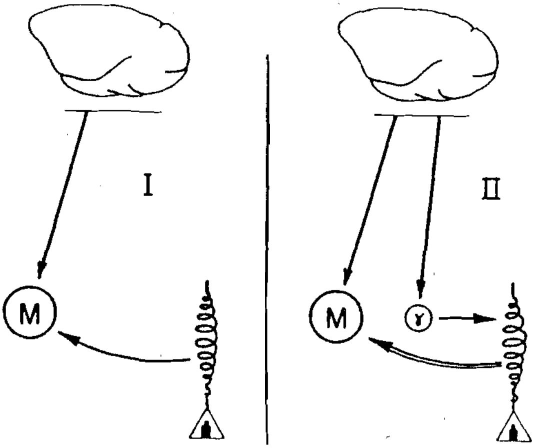

Figure 3. Role of spindle mechanism in integrating peripheral and cortical influences at the spinal level. In circuit I the motor neuron has the total responsibility. In circuit II the spindle, through its motor innervation, takes over an important part of the integrating activity. From Mortimer and Akert 1961. p. 246.

If this idea were correct and the motor nerves to the spindle were cut, movement should be drastically affected. Two graduate students, G. Shambes and J.L. Smith, used a local anaesthetic infiltration to differentially block spindle motor nerves while leaving motor nerves to contractile muscle intact, first into the popliteal space, bilaterally, (Shambes, 1969), and second into the axilla where the radial nerve lies outside the major sheath of the brachial plexus (Smith, Roberts, & Adkins, 1972). We expected the subjects to fall over with the popliteal space block: wrong track again! The deficit was so slight it was difficult to measure. Essentially all we could detect was a slight increase in postural sway. I myself was a pilot subject. Although it felt odd, I found no apparent difficulty in walking or jumping (as well as one can jump in an operating room).

With the radial nerve infiltration experiments which blocked gamma motor nerves to spindles in the extensors of the elbow, wrist and fingers, more discriminating measures were used, including a "finger-to-nose touch" and a dart throw (Smith, et. al, 1972). With the "finger-to-nose" touch the most obvious deficit was failure of deceleration. Instead of touching the nose gently, subjects banged into their faces and sometimes missed their noses. Surprise registered on their faces when things went wrong. In the dart throw, subjects used their normal timing but their hand with the dart had not accelerated to the needed speed when they released so the dart fell far short of the target. Again the film registered the surprise on their faces when they saw what happened. Apparently they were not fully aware of the deficiency until something untoward occurred. They corrected quite quickly by using more shoulder action to help "throw" the elbow into extension.

These studies indicated that we needed to know more about movement accelerations and timing. Available details were rather sketchy, with Ruth Glassow's being among the few available. It seemed that more information was needed in order to postulate control mechanisms. In the meantime Betty Atwater had been looking into how to teach college women to throw. The efforts were proving to be singularly unsuccessful. Betty decided she needed more detailed information about the movement. This lead to her Ph.D. study which may be familiar to many (Atwater, 1970; 1979). One of her intriguing findings related to the path of the ball in the hand of a subject throwing for speed. The ball did not increase its speed gradually as would be expected if the performer was trying to make use of leverage principles (see Figure 4a). Rather it moved relatively slowly

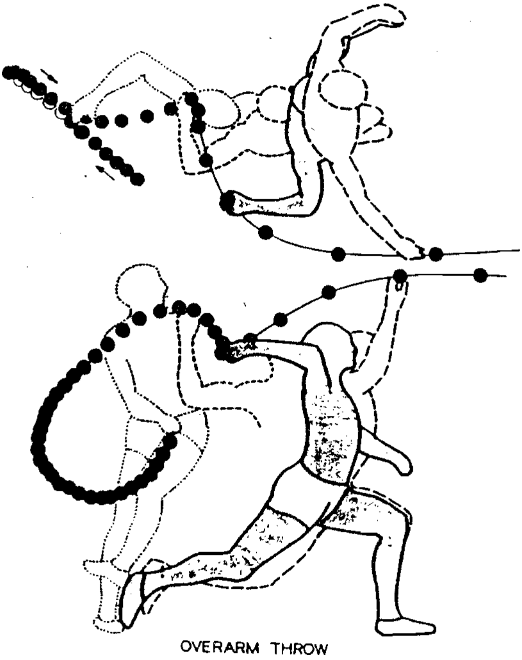

Figure 4a. Path of Ball in hand of skilled performer during execution of overarm throw for speed. Top view above, side view below with interval between ball positions or 14.8 and 11.7 ms respectively. Shaded tracings indicate comparable points in time. Ball slows down in space then accelerates to release velocity. From Roberts & Metcalf, 1968, p. 318.

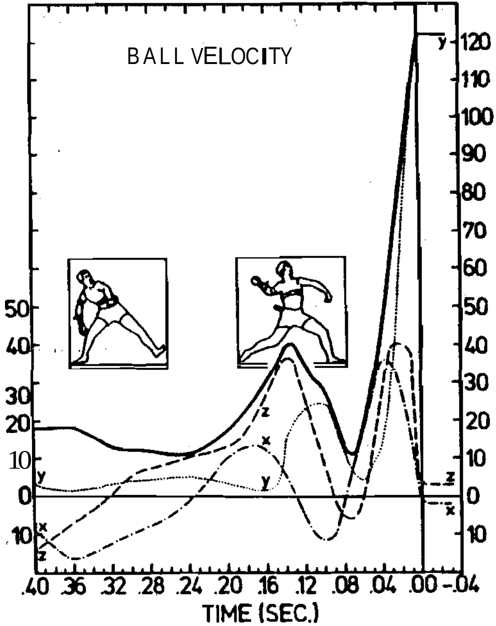

Figure 4b. Component and resultant velocity of ball in hand **during** final 400 ms before release in a skilled **overarm** throw for speed. Solid line, resultant velocity; x, **lateral** component: y. **anterior-**posterior component: **z,** vertical component. **Tracings** of subject at 2% sec and .130 sec. Early increase in resultant velocity produced primarily by vertical (z) component and mainly associated with right shoulder abduction and start of shoulder external rotation. **Sharp decrease in velocity** associated with shoulder **external** rotation as forearm rotates beyond vertical. Ball then **accelerates from 11 ft/s to 122 ft/s in - 75** ms. Adapted from **Atwater,** 1970, pp. 200-222.

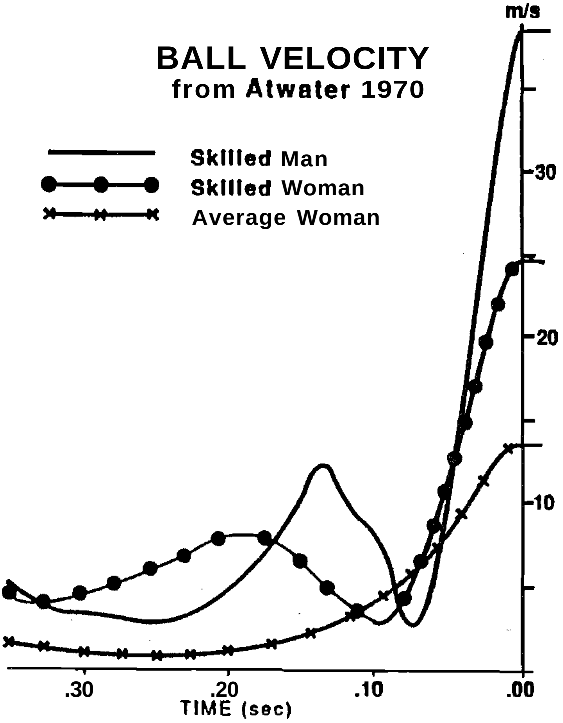

Figure 4c. Resultant velocity of ball in hand during final 400 ms in overarm throws for speed. Three different performers. Least skilled **performer shows** lower velocities throughout. with final acceleration (slope **of line**) to release being least and starting earliest in time. (skilled performer **also** shown in Fig. 2) Adapted from **Atwater,** 1970 p. 188.

while the body performed most of its actions including striding, weight transfer and trunk rotations, and only picked up speed near the end (see Figure 4b). The whole motion could last 1 to 15 sec, but the final acceleration could take as little as 75 msec. i.e., after about 90-95% of the performance the ball could be moving as slowly as 3 m/s. In the remaining 5-10% of the time it would accelerate from 3 m/s to 38 m/s. That is a phenomenal performance. In addition, Atwater recorded that the trunk and hip rotations were decelerating before release? As already mentioned, this result does not seem compatible with leverage principles. Thus, if the performer is not relying primarily on leverage principles to achieve high velocity values, how is he/she doing it? Of course, not all performers achieve high values as can be seen in (Figure 4c). What are the less skilled not doing from a mechanical point of view and how can we help them reach a higher performance level?

I can remember one of our anatomy professors asking Betty at the time of her Ph.D. orals what muscles she thought could be providing such high ball accelerations (on the order of 465 m/s2). She. as well as the rest of us, was unable to come up with a likely answer. It didn't seem to us that triceps was a very good candidate. Medial rotators didn't seem very likely either especially since Betty's films seemed to show medial rotation being delayed until about the last 15-20 msec before release (see Figure 5).

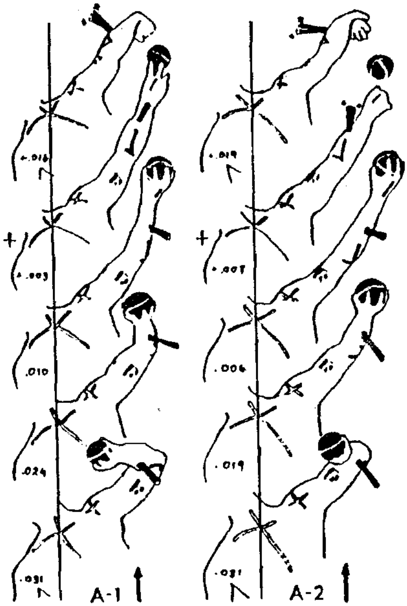

Figure 5. Back view of two skilled male performers during the final - 31 ms before release and - 20 ms into "follow through" in overarm throws for speed: Forearm pronation and internal rotation at shoulder occur late. Adapted from Atwater, 1970 p. 227.

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[2] Atwater was not satisfied with the detailed accuracy of her measures because she suspected that the projecting attachments she used for measurement rotated independently because of their inertia.

We began to wonder if some centripetal force or segment interaction forces were involved. While we had nerve block studies underway, we tried complete paralysis of the elbow extensors. As Dobbins reported (in Roberts, 1971), we found that the elbow would in fact extend in the throw without elbow extensor muscle action (see Figure 6). However, it also flexed much too far without the elbow extensors to hold it near 90° (we hadn't thought about that!). As with the dart throw experiment the subject used his "normal" timing in the first throw after nerve block, so that by the time he released the ball the elbow had not extended far enough, and the ball velocity was down by 55%. By the 6th throw, he had managed to manipulate his movement so that the speed was back up to 81% of normal. Elbow extension at release had increased considerably. The data showed that he had increased trunk lateral lean which, we thought, could help to "throw" the forearm into extension, but it also showed that he had reduced biceps activity so that the elbow did not flex as far in the first place.

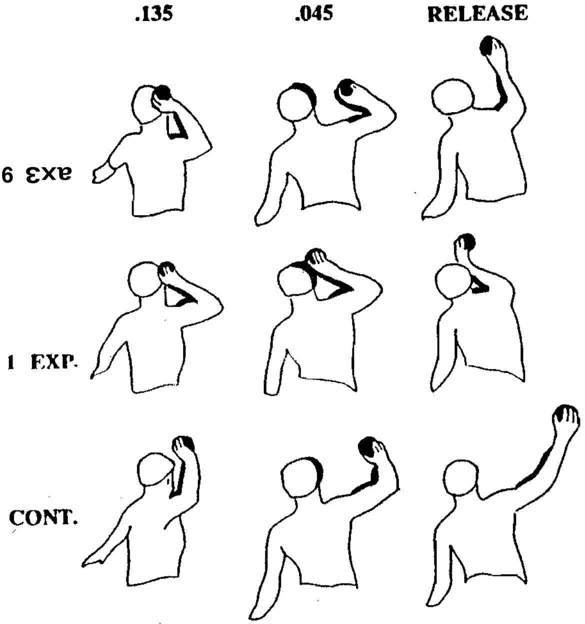

Figure 6. Effect of local anesthetic block of radial nerve to elbow and wrist extensor muscles on an overarm throw for speed. Back views at .135 sec and .045 sec before release and at release. Control throw, bottom, 1st experimental throw following paralysis, middle; 6th experimental throw, with extensor muscles still paralysed. Elbow angle, and proximity of ball to head at .045 sec., are darkened for emphasis. In 1st exp. throw elbow flexes much too far when extensors are paralysed, but still extends to some degree before release. In 6th exp. throw subject has made adaptations so that elbow does not flex as far and extends further before release. Adapted from Dobbins 1970 research report.

These indications that intersegmental forces, not just muscle forces, could be involved in movement execution led to the pursuit of methods for determining accelerations which would lead to forces. Curve fitting was a serious problem here (e.g. Zernicke, et. al, 1976; Phillips & Roberts, 1983; Woltring 1985) but I will not go down that track right now. Some puzzling, or perhaps, enlightening things began to emerge. One was the kinematic observation that peak accelerations tended to occur around reversal from backswing to forward swing (see Figure 7) not primarily during forward swing as we had thought, (another wrong track!). Gowitzke (1975) looked at this finding more closely, examining conditions that might affect peak accelerations in an underarm motion for maximum velocity, with the trunk essentially stationary, varying range and speed of backswing. It was interesting to observe that when subjects were commanded to use a "fast" backswing as opposed to their preferred speed of backswing, peak acceleration tended to occur before reversal suggesting that some acceleration was wasted in stopping the backswing rather than being available to speed up the forward swing.

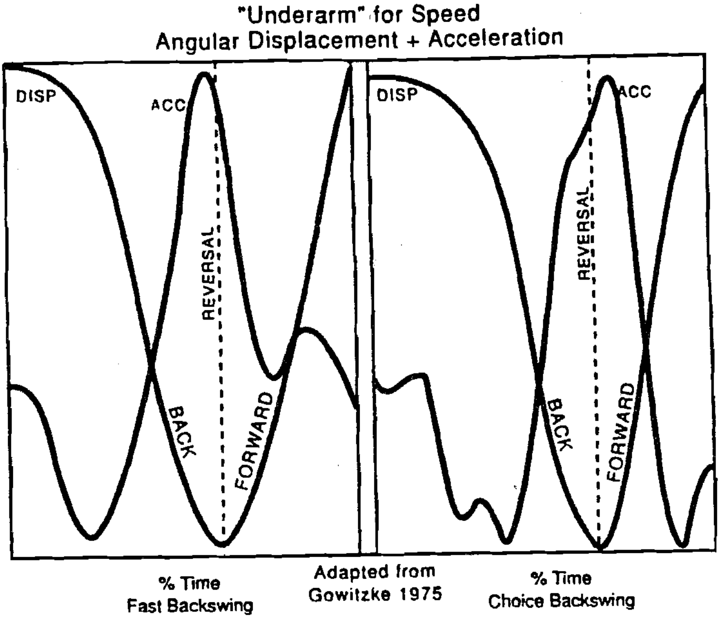

Figure 7. Displacement and derived acceleration of upper extremity in an "underarm" motion for maximum velocity: measured by potentrometer on a mechanical arm aligned with and comfortably connected to the performer's limb at the wrist. Left, subject instructed to execute a fast backswing: acceleration peaks before reversal. Right, subject chooses backswing speed: acceleration peaks after reversal for forward speed development. Adapted from Gowitzke, 1975.

Another interesting kinematic observation was that, in a toe kick for speed, the thigh negatively accelerated, slowing down before impact, a phenomena similar to that of the trunk in throwing. In addition, the positive acceleration of extension at the knee joint was almost directly opposite the negative acceleration of the thigh (see Figure 8, Roberts & Phillips, 1977).

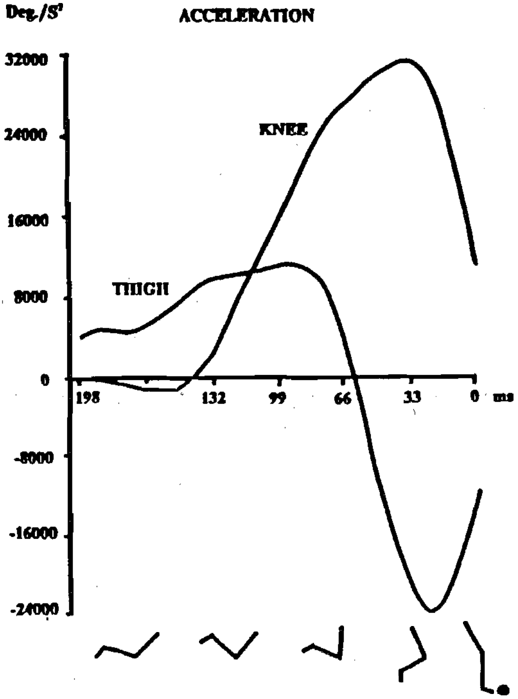

Figure 8. Soccer toe kick. Angular acceleration of thigh segment and intersegmental knee angle. Peak negative acceleration of thigh and positive acceleration of knee are almost opposite. From Roberts & Phillips, 1978.

This rather unexpected event, while confirming the inadequacy of leverage as a prime factor in speed development, suggested that proximal deceleration might be linked to distal acceleration in some way. Zemicke's (1974) kinetic data on the toe kick added a dimension by indicating that the knee muscular moment tending to cause extension of the shank dropped off toward zero well before ball impact. yet angular acceleration of the shank was still present beyond this point (see Figure 9).

What force was producing the later phase of the acceleration? Sally Phillips and I thought perhaps the interacting joint forces at the knee over and above the knee muscular moment might be involved (Phillips. Roberts. & Huang, 1983). We tried to demonstrate their influence by mathematically eliminating the knee muscular moment at various times throughout the swing phase of a run and a toe kick. The simulation showed that the shank could be accelerated and decelerated through its joint connection to the thigh without any muscular moment (see Figure 10). Sally and I interpreted this to mean that thigh deceleration facilitated knee extension. (Plagenhoeff, [1971] had

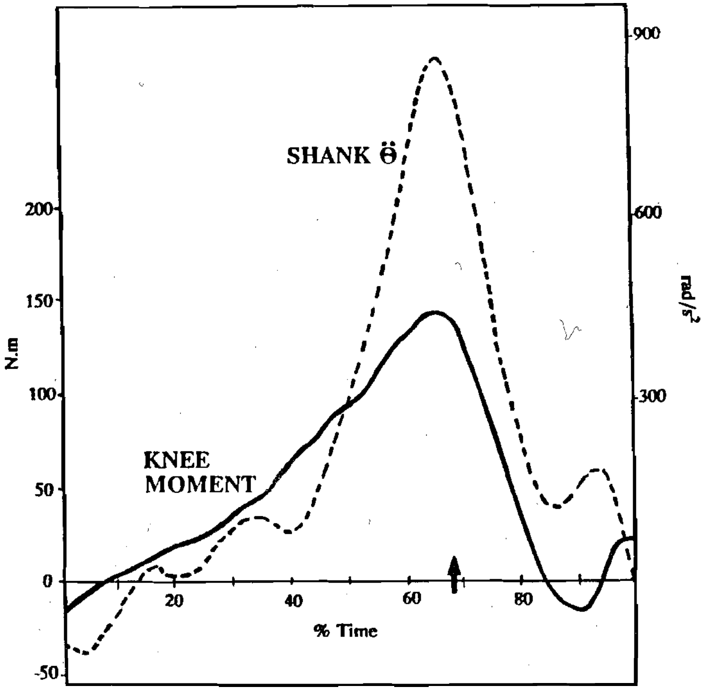

Figure 9. Soccer toe kick. Knee muscular moment of force and shank angular acceleration Positive shank acceleration continues after muscular moment has become negative. Adapted from Zernicke, 1974.

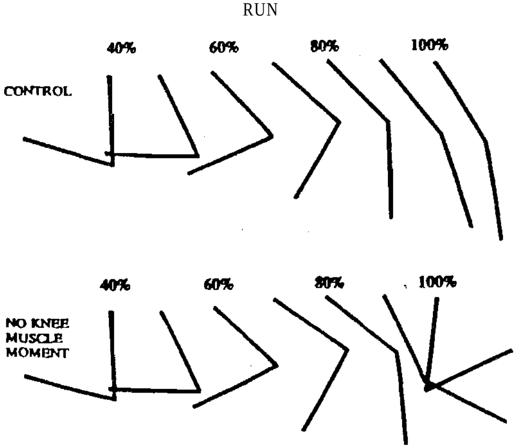

Figure 10. Normal motion of recovery limb in running (above) and motion mathematically predicted without knee muscular moments (below): 40% - 100% of swing. Subject 64 years. Motions are similar 40% - 70% when knee moment is near zero. After 70%, with knee flexor moment gone, thigh rotates backward too far and knee extends beyond anatomical limit. Adapted from Philips & Roberts, 1980 p. 269.

made such a suggestion earlier). However, Putnam (1983) made the reverse interpretation in her study of punting, namely that the shank's angular motion decreased thigh angular velocity. Both groups found that the effects produced depended on where the segments were in relation to one another at the time that "motion dependent moments" occurred. Putnam (1980, p. 77) conceded that if the knee had extended beyond 90° the "motion dependent" thigh deceleration moment could assist late knee extension, (but in her simulation such an occurrence did not produce optimal foot speed). Similarly Phillips (1978) had found some positive contribution of the moment of intersegmental knee joint forces to the rate of change of momentum of the shank (I) in the last 50 msec of a toe kick (see Figure 11) when the knee was reversing and developing extension velocity (p. 281). So it seemed that intersegmental forces might play some direct contributing role under normal circumstances, but it appeared to be a rather minor role.

While Phillips et. al (1983) and Putnam (1983) applied the same Newtonian equations in treating their data, they presented the data in different forms. Phillips retained the basic form with moments of force calculated about the center of mass. Putnam moved the center of rotation to the proximal joint and expressed all moments as functions of kinematic variables. The magnitude and direction of moments. especially moments due to segment interactions, are different in the two procedures; hence, presentations and interpretations can be different The differences in calculated moments are related partly to the fact that the linear acceleration of the center of mass. as well as the mass moment of inertia about the center of mass, are different from those of the proximal end. Other investigators have used still other rearrangements and

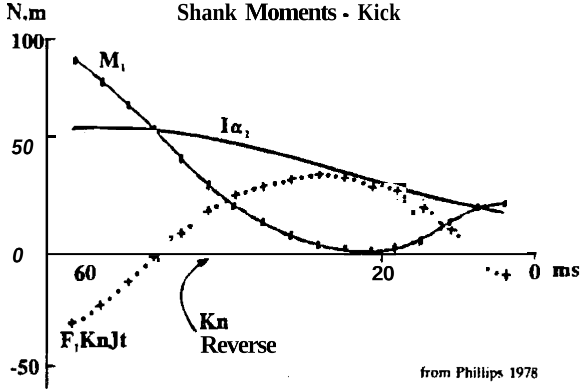

Figure 11. Soccer toe kick. Ball impact at 0 ms. M_1, knee muscular moment of force. F_1Kn.Jt, moment of knee joint forces about shank center of mass. I, resulting rate of change of angular momentum of shank. Kn. reverse, time of knee reversal from flexion to extension. Joint force moment becomes positive before knee reversal. and is in the same direction as muscle moment Adapted from Phillips 1978. p. 281.

presentations of data (e.g. Chapman, et. al, 1984; Felmer & Dapena, 1989; Hong, 1991); thus, the waters are rather muddied. Recently, Putnam (1991) has explained the 2D effect of each calculated moment in her equations on leg segments (5 for the shank and 7 for the thigh) viewing each independently. She has warned, however, that in reality they are not independent and that an alteration in one kinematic parameter or resultant moment ("muscle" moment) of either segment affects all other moments, thus interpretation"can be confusing."@.144). With these limitations in mind let me present to you some of the current material on throwing.

Felmer and Dapena (1986,1989) and Feltner (1989) have fairly recently tackled the kinetics of the throwing arm in 3D, Bullard (1989) has worked on 3D trunk kinematics, and Hong is currently including 3D trunk kinetics in his study of pitchers. Figure 12, adapted from Feltner (1989), roughly illustrates in 2D the last 120 msec of the throwing motion (remember the movement is actually 3D so there is distortion in Figure 12). The line segments represent the forearm, upper arm, and trunk.

It can be seen that the angular motion of the trunk segment is slowing down as release is approached, as mentioned earlier, and that horizontal motion of the humerus in relation to the trunk is very limited, so that the trunk motion is responsible for most of the humerus motion in this plane of space. Thus the linear motion of the shoulder and elbow are largely due to the trunk. Shades of the long lever principle! The hand and ball are also moved by the trunk but not as much since elbow flexion and external rotation of the humerus tend to move them in the opposite direction. (Motion of the shoulder

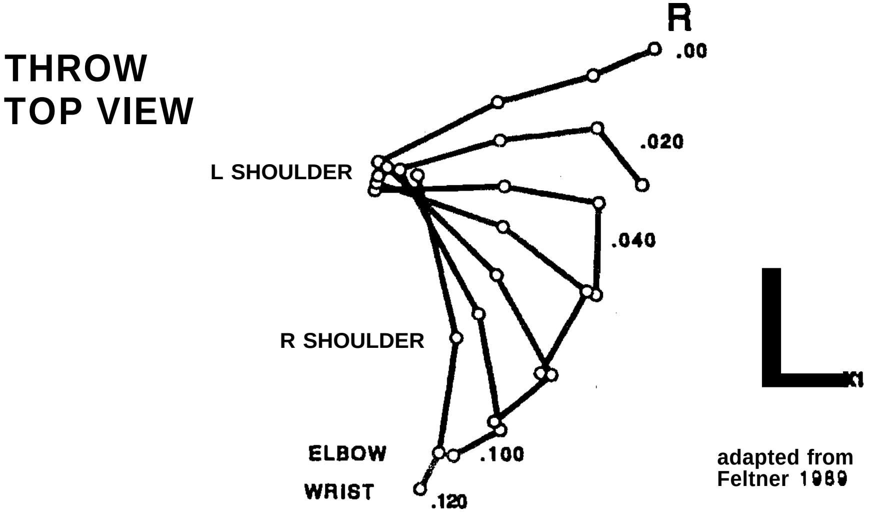

Figure 12. Sequence showing overhead view (Y_2 vs X_2) of the shoulders and throwing arm of a baseball pitcher. R.. release. Throw is in the Y_1 direction. Images at .02 sec intervals. Trunk rotation slows before release. Change in angle between trunk and upper arm is small. Adapted from Felmer 1989. p.441.

girdle which may be very important is not included). Extension of the elbow does not begin until somewhere around 60 msec before release. Atwater (1979) showed essentially the same kinematics in her earlier studies.

Di An Hong's (1991) representation of the hips and upper trunk is shown in Figure 13, reconstructed from his digitized smoothed coordinates. The darkest surface. representing the right, throwing arm side of each block shows the hips already rotating forward near stride foot contact. 192 ms before release, with the upper trunk lagging behind, then starting to catch up. The rotations slow before release but the trunk can be seen to lean to the left and forward, with the hips tilting mainly forward. Since the humerus does not abduct much during this time and horizontally adducts only a small amount (Hong 1991) the trunk motion, as mentioned above, accounts for most of the linear velocity of the shoulder and elbow. Figure 14 shows the resultant linear velocities. together with that of the hand and ball, for a pitcher, from the time of stride foot contact (SFC) until release, a duration of –.2 sec (Hong 1991). It can be seen that the elbow moves faster than the hand and ball from about 125 ms until elbow extension (EE) starts at 64 ms mainly because flexion of the elbow and lateral rotation of the humerus delay the hand and ball. Pronation/supination and shoulder girdle motion, though both very important, could not be included in this study.

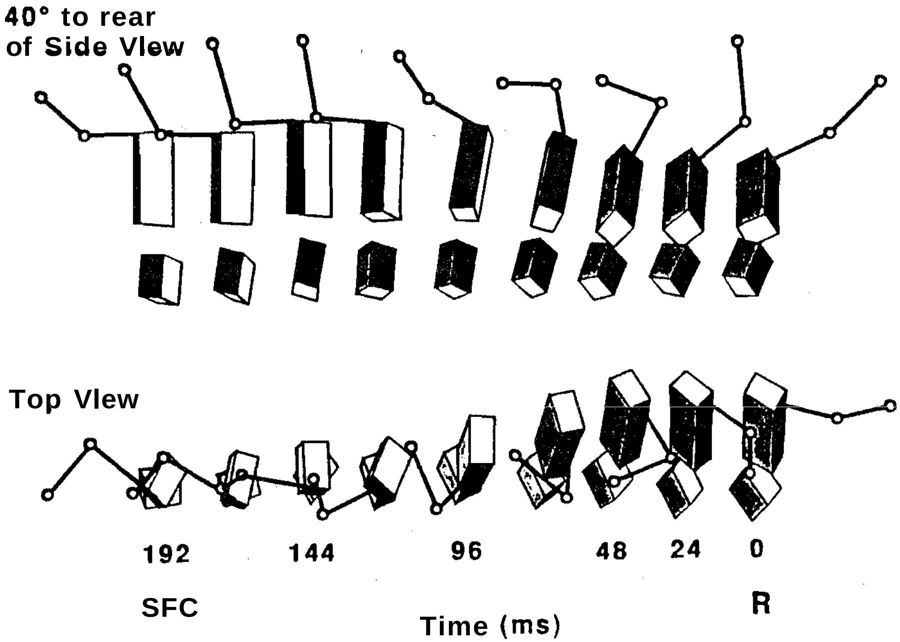

Figure 13. Model of hips. upper trunk and throwing arm reconstructed from 3D film coordinates of a baseball pitch. From Hong. 1991.

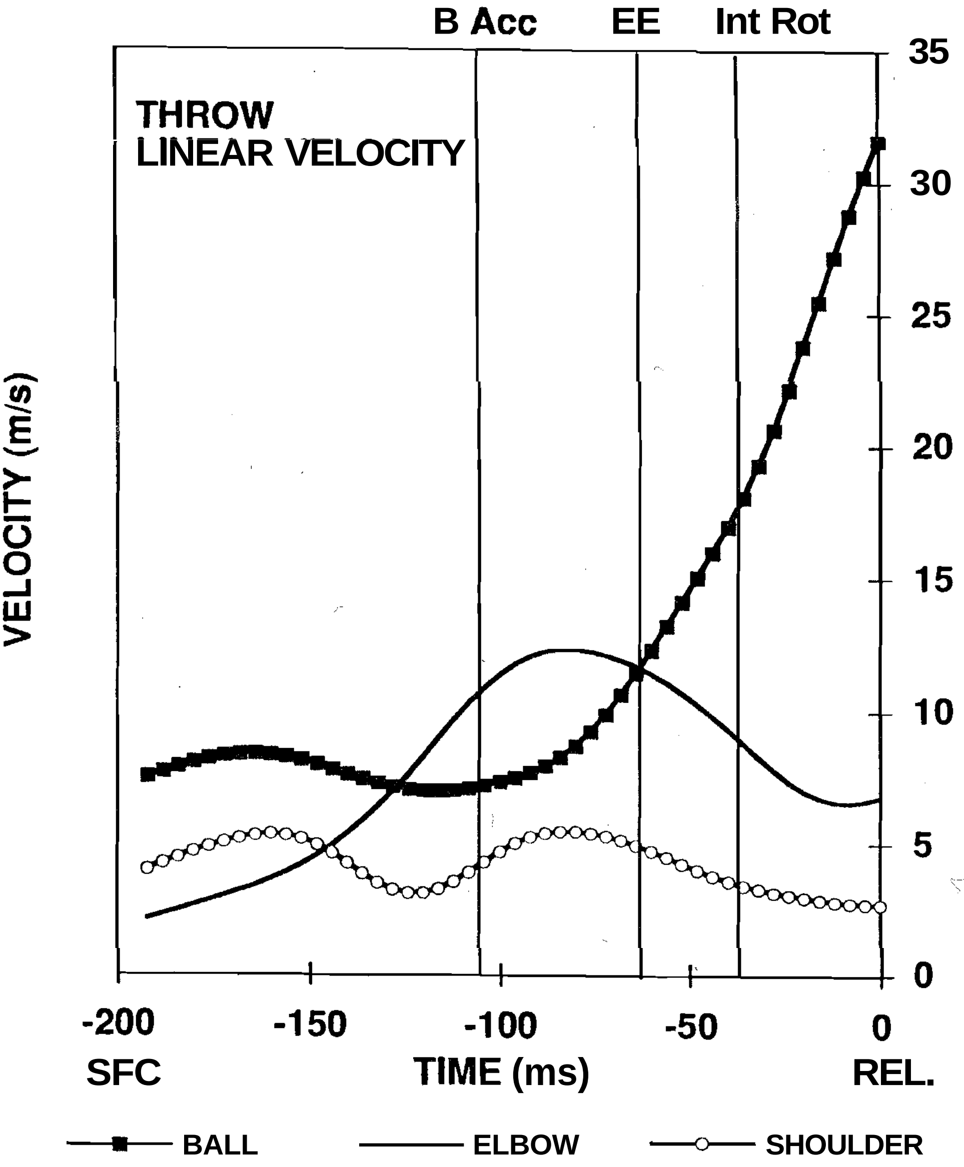

Figure 14. Resultant 3D linear velocity of shoulder, elbow and ball in hand of pitcher. SFC. stride foot contact REL.. release. E.E. start of elbow extension. INT ROT, start of internal rotation of humerus. Velocities are not necessarily in the same direction. From Hong, 1991.

In making 3D calculations of the moments of force involved in pitching, Feltner (1989) and Hong (1991) have used somewhat different procedures in obtaining angular velocities and accelerations and in presenting and interpreting data on the throwing arm. Hong, whose data are just coming out of the computer, is using a six segment model which includes the trunk and contralateral (left) arm as well as the throwing (right) arm. In presenting the data he has chosen to take moments of force about the proximal joint and to show the influence of a proximal segment on a distal one by way of the linear acceleration at the connecting joint. Since this value includes the effect of muscular forces as well as intersegmental forces, the latter cannot be isolated. The data may, however, give some additional insight into segment interactions in 3D throwing.

The matrix form of the equations of motion used by Hong is:

$$F_{jp} + F_{jd} + G_j = m_j X_j$$

$$r_{jd} F_{jd} + M_{jp} + M_{jd} - r_{jc} m_j X_{jp} + r_{jc} G_j' = I_{jc} w_j + \underline{w_j I_{jc}} w_j$$

where underlining denotes a square matrix, F_p and F_d are proximal and distal forces respectively, G and M are segment weight and mass respectively, X and X_p are linear acceleration of the center of mass and proximal end respectively, M_p and M_d are muscular moments of force about the proximal and distal ends respectively, r_d is the length of the segment, r_p is the distance from the center of mass to the proximal end, I_p is the mass moment of inertia about the proximal end and and are angular acceleration and angular velocity respectively.

I will briefly illustrate some of Hong's preliminary findings with three examples from one throwing trial[3]. First, and perhaps most complex, is the upper trunk.

Figure 15 illustrates rotation moments[4] (a), and resulting rate of change of momentum (b), about an axis aligned along the spine, from the time of stride foot contact (SFC) until release (REL.). The axis moves with the trunk. The proximal muscle moment (MMus.P), which represents trunk rotators, is tending to rotate the trunk forward (CCW) immediately after stride foot contact (SFC). In this trial there is a pause in the moment probably because the angular velocity of rotation is almost constant, and then it picks up again, perhaps in part because of the load created by a small amount of horizontal adduction of the humerus about this time (see Figure 12), together with the forearm being externally rotated past the vertical. Shortly before elbow extension (EE) begins, the muscle moment reverses direction, tending to slow trunk rotation (CW). However, as internal rotation of the humerus gets underway, the muscle moment again

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[3] This trial may or may not prove to be representative of other trials and subjects.

[4] For simplicity distal moments at the left shoulder are not shown. Proximal acceleration moments are too small to show.

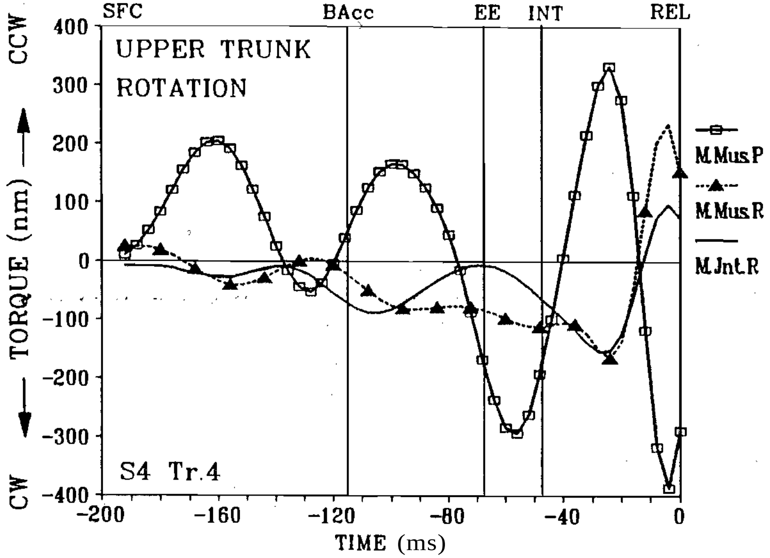

Figure 15a. Baseball throw. Upper trunk rotation. Axis aligned with, and moving with spine. **M.Mus.P,** moment of proximal musculature. **M.Mus.R,** moment of distal musculature at right shoulder pint. **M.Jnt.R.,** moment of joint forces at right shoulder. Distal moments at left shoulder not shown. See text. From Hong. 1991.

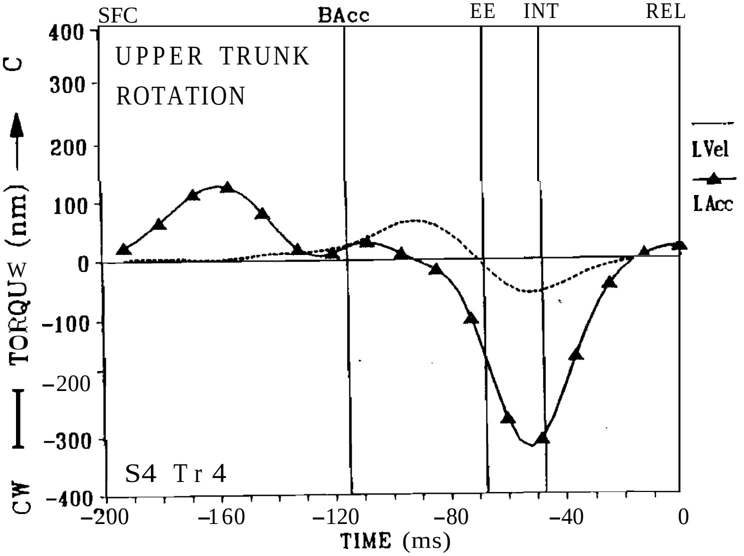

Figure 15b. Rate of change of angular momentum of upper trunk. **I.Acc.,** angular acceleration component. **I.Vel.,** angular velocity component. See text. From Hong. 1991.

becomes positive (CCW) seemingly to resist this additional load. It is difficult to know if the 2nd negative moment is real or an artifact related to end point conditions. (e.g. Phillips & Roberts 1983).

The influence of the proximal linear acceleration of the point in the hips about which trunk rotation moments were modeled has only a small component in this plane (not shown therefore); hence, the muscular moment (M.Mus.P) accounts for much of the angular acceleration component of the rate of change of trunk momentum (I.Acc.), (Figure 15b). The angular velocity component of the rate of change of momentum (I.Vel.), which is related to angular velocities in directions other than that of the segment, is relatively small.

The left arm has a small effect on the trunk (not shown) which, in general, assisted with trunk rotation early but not later. At the right shoulder (Fig. 15a) both the distal muscular moment (M.Mus.R) and distal joint force moment (M.Jnt.R) acting on the trunk tend to hold the trunk back while attempting to keep the humerus in line with the shoulders.

Since the humerus moves with the trunk for the most part (see Figures 12 & 13) except in external and internal rotation, I will only show the latter moments acting on the humerus (see Figure 16). They are relatively uncomplicated by anything but the load of the forearm.[5]

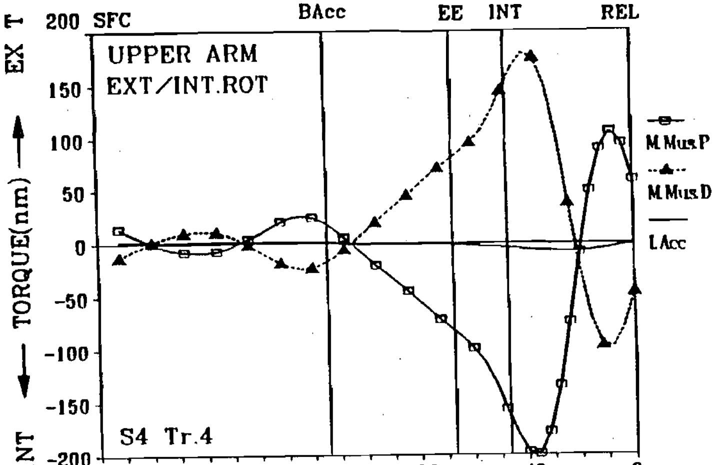

Figure 16. Baseball throw. Internal/external rotation of Humerus. M.Mus.P., moment of proximal musculature. M.Mus.D., moment of distal musculature. I.Acc., acceleration component of rate of change of angular momentum: velocity component too small to show. See text. From Hong, 1991.

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[5] Hong measured internal/external rotation of the humerus from the middle of the wrist to the shoulder joint.

This opposing load is represented by the "distal muscle moment" (M.Mus.D) which, in this case, may be due to many structures around the elbow joint. The rotators of the humerus (M.Mus.P) provide a small external rotation impetus between − 145 and − 105 ms before release to bring the forearm up toward vertical. They then immediately build an internal rotation moment, to the slow down the ongoing external rotation. Peak internal moment is reached just after the reversal from external to internal rotation (INT), which should provide a high internal rotation acceleration to add speed to the ball (see Figure 14). The small blip of external rotator moment just before release may be in reaction to late pronation of the forearm as elbow extension (EE) moves and the long axis of the forearm closer to that of the humerus. The acceleration component of the rate of change of angular momentum of the humerus (I.Acc) is small because the moment of inertia about the long axis is small. The angular velocity component was too small to show.

For the illustration of flexion/extension moments on the forearm (see Figure 17) the axis is aligned with the humerus when it is in 90° abduction and the forearm is flexed 90°. When the humerus externally rotates, carrying the forearm with it, elbow extension gets closer to the plane of trunk rotation and early elbow extension is in the throwing direction. When internal rotation gets under way, however, elbow extension contributes less and less to ball velocity.

The two moments on the forearm (plus hand and ball) in the flexion/extension direction shown in Figure 17a are the muscle moment (M.Mus.P) and the moment produced by the proximal linear acceleration (MAP) in the relevant direction. The latter, which includes the effects of segment interactions. is directed along the long axis of the humerus and is presumably caused partly by the lateral bending of the trunk to the non-throwing side. Both moments are slightly extensor at stride foot contact (SFC), but the muscle moment becomes flexor briefly, working to increase flexion (see Figure 13) against the tendency of the proximal segments (MAP) to slow flexion. It then pins the proximal acceleration moment in slowing flexion until the reversal point (EE), when both accelerate extension. These data agree quite well with the earlier nerve block dam when the extensor muscles were paralyzed (see Figure 6). In that case the forearm could flex too far because the flexor muscles were still intact It could still extend, however, because the intersegmental moment (MAP) would have been extensor, even without the extensor muscles.

The elbow muscle moment (M.Mus.P) becomes flexor shortly after internal rotation of the humerus begins, presumably to slow elbow extension before it approaches the end of its range, where it can contribute less and less to ball velocity. Notice that it is the intersegmental moment (MAP) that tends to cause elbow extension to continue and, if unopposed, could damage the joint. Moments near release do not need to be large because the velocity component of rate of change of momentum (I.Vel.) is largely balanced by the acceleration component (I.Acc., see Figure 17b).

Of the examples shown from Hong's study, elbow flexion/extension is the one that

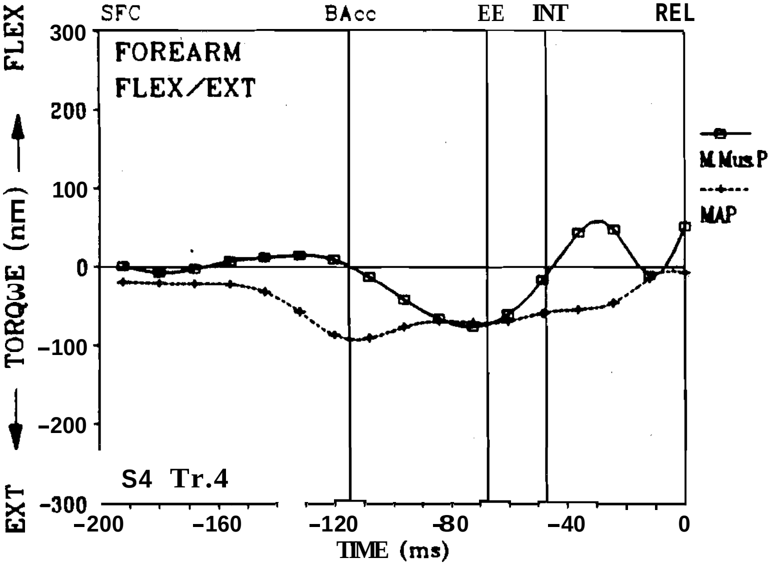

Figure 17a. Baseball throw. Flexion/extension of forearm hand and ball. M.Mus.P., moment of proximal musculature. MAP moment due to linear acceleration of proximal (elbow) joint. See text. From Hong, 1991.

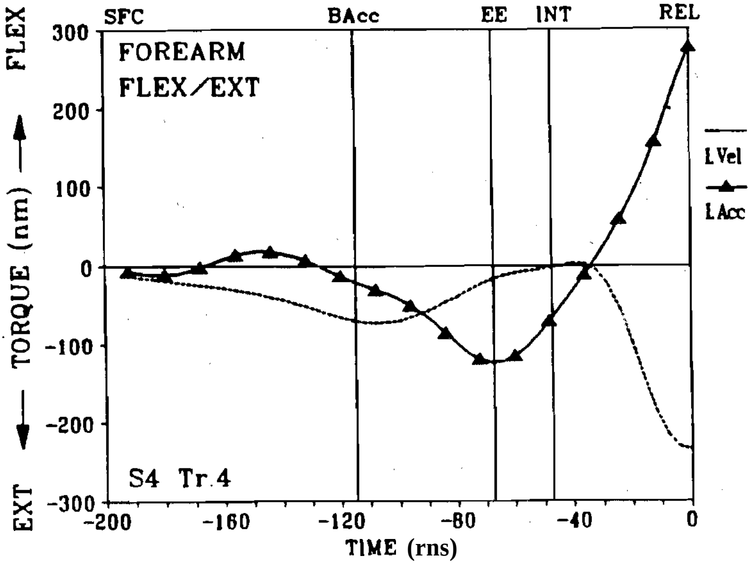

Figure 17b. Rate of change of angular momentum of forearm, hand and ball. I.Acc., angular acceleration component. I.Vel. angular velocity component. See text. From Hong, 1991.

provides information on the supporting role of segment interactions (MAP) in throwing. Our earlier studies (e.g. Atwater, 1970; Dobbins, 1970) indicated that interactive forces had the capability of playing a supporting role in velocity development. The recent studies confirm that they actually do and help to define[6] how big a role and when it is played. In addition, these recent studies may help to specify when and to what extent intersegmental forces could be detrimental to performance and potentially damaging to joints if not controlled by muscle activation.

In summary, those findings that seem to be on the right track for velocity development include the following: Cortical control of muscle spindles plays a role in precise initiation of muscle activation. Precise muscle activation generates accurate forces that accelerate body segments. Semi-rigid anatomical segments conform to leverage principles which, therefore, play a role in velocity development. Leverage principles alone, however, are not a sufficient explanation of velocity development. Accelerating and decelerating segments, mechanically linked to other segments, generate reaction forces in the linked segments. Reaction forces are also involved in velocity development. When certain distal muscles are paralyzed, either in fact or mathematically, some velocity can still be developed but apparently not the target velocity. Segment interaction forces are not a sufficient explanation of velocity development either: muscular forces are essential. Recent findings indicate that skilled performers supplement precisely activated muscle forces with intersegmental forces to achieve the highest resultant velocities.

Many years and many wrong tracks were followed in getting this far along what I hope is now a right track in understanding velocity in motion. The track still stretches far ahead, though, so there is much to be done by others. I cannot wait to find out where it leads!

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[6] The reader is reminded that the calculated intersegmental moments of force are not purely passive bone on bone forces but can include components from muscle and joint structures.

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