Differentiating Ability in Users of the ReWalk™ Powered Exoskeleton

An Analysis of Walking Kinematics

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***Abstract*—The ReWalk™ powered exoskeleton assists thoracic level motor complete spinal cord injury patients who are paralyzed to walk again with an independent, functional, upright, reciprocating gait. We completed an evaluation of twelve such individuals with promising results. All subjects met basic criteria to be able to use the ReWalk™ – including items such as sufficient bone mineral density, leg passive range of motion, strength, body size and weight limits. All subjects received approximately the same number of training sessions. However there was a wide distribution in walking ability. Walking velocities ranged from under 0.1m/s to approximately 0.5m/s. This variability was not completely explained by injury level The remaining sources of that variability are not clear at present. This paper reports our preliminary analysis into how the walking kinematics differed across the subjects – as a first step to understand the possible contribution to the velocity range and determine if the subjects who did not walk as well could be taught to improve by mimicking the better walkers.**

Keywords—ReWalk™, powered exoskeleton, Spinal Cord Injury, walking performance, training methods.

I. Introduction

Spinal cord injury currently affects function of approximately 200,000 Americans [1], with nearly 11,000 new injuries annually in the United States alone [2]. Mobility limitations are a key factor contributing to reduced function, health and life satisfaction in this population [3, 4]. Non-recreational walking accounts for a significant fraction of activity for the average adult. Yet, for thoracic level and higher motor complete injuries upright bipedal walking is highly impractical, if at all, achievable.

Most individuals with a motor complete thoracic level SCI are forced to rely on manual wheelchair propulsion for locomotion [5]. While this does allow effective movement around the house, at work and in the community, there are a number of disadvantages and limitations associated with this mode of locomotion [6]. Environments must be wheelchair accessible. Conventional wheelchairs do not load the legs [7] which is a key factor to maintaining bone strength. Excessive sitting can lead to pressure sores [8, 9] and promote joint contractures. Using arms for propulsion can result in upper limb joint overuse syndrome [10], [11] as well as neuropathy. Last, but not least, wheelchairs also do not allow for eye level interaction with able bodied adults.

The other locomotion options available to this population are more cumbersome and less efficient than the wheelchair. Functional Electric Stimulation (FES) systems have only been successful for very limited distance ambulation [12] due in part to quick onset muscle fatigue and high energy expenditure. Knee-ankle-foot orthoses (KAFOs) are heavy, cumbersome, and difficult to apply. Reciprocal gait orthoses (RGO), hip guidance and isocentric orthoses (IRGO) offer increased ambulation over conventional KAFOs, but for short distances only. Their use is still limited by high energy demands, potential for increase upper limb overuse [13], difficulty donning/doffing, not fully independent use, and brace mechanical integrity. These braces are also difficult to use in other important activities that contribute to overall mobility independence, for example sit to stand or getting in and out of a car [14],[15].

Powered exoskeletons have the potential to address some of these limitations. While they have been around for many decades, the technology seems to have experienced resurgence recently. There are currently two devices in the US that can be used for upright untethered walking in paralyzed individuals – the EKSO (Ekso Bionics, Berkley CA, USA) and ReWalk™ (Marlboro MA, USA and Yokneam Isreal). Neither yet allow the speed of a wheelchair on flat level ground. However both can navigate a wider range of terrains than a wheelchair, while allowing the advantage of being upright. Initial energy cost to use is high in the ReWalk™ but it does not appear prohibitive to purposeful community ambulation for an experienced user.

A. *What is ReWalk™?*

The ReWalk™ is a powered exoskeleton that allows thoracic motor complete individuals with SCI to walk independently (Figure 1). ReWalk™ contains a pair of hip and a pair of knee joint motors powered by rechargeable batteries and a control system housed in a user-worn backpack. The system is entirely self-contained and subject-directed. Users control their own walking through minor trunk movements and

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a wrist-pad controller. A tilt sensor determines the trunk angle and generates a prescribed hip and knee displacement (angle and time) that results in a step.

The ankles use a simple double action orthotic joint with limited motion and spring assisted dorsiflexion adjustable through screw tension. Velcro straps, shoes and a waist belt are used to secure users in the device. Forearm crutches are mandatory to achieve stability when upright as subjects have little or no voluntary trunk control with this level of injury. Subjects can interact with the system through a wrist-watch style controller. Via this controller, the subject can activate the stand, sit or walking mode programs. In the walking program, the software interprets torso tilt sensor data and generates alternating limb coordinated motion to produce bipedal walking. The system is coded to prevent two sequential steps of the same leg. Joint angle displacements for the knee and hip can be adjusted using an external computer to optimize the walking characteristics or implement a training mode. A manual mode of operation can be used to trigger steps bypassing the tilt sensor. The same mode of operation can be used to trigger sit-stand-sit transfers. ReWalk^{TM} is suitable for adults who have preserved bilateral upper extremity function as well as the capacity for assisted standing (such as in a standing frame or with braces and crutches).

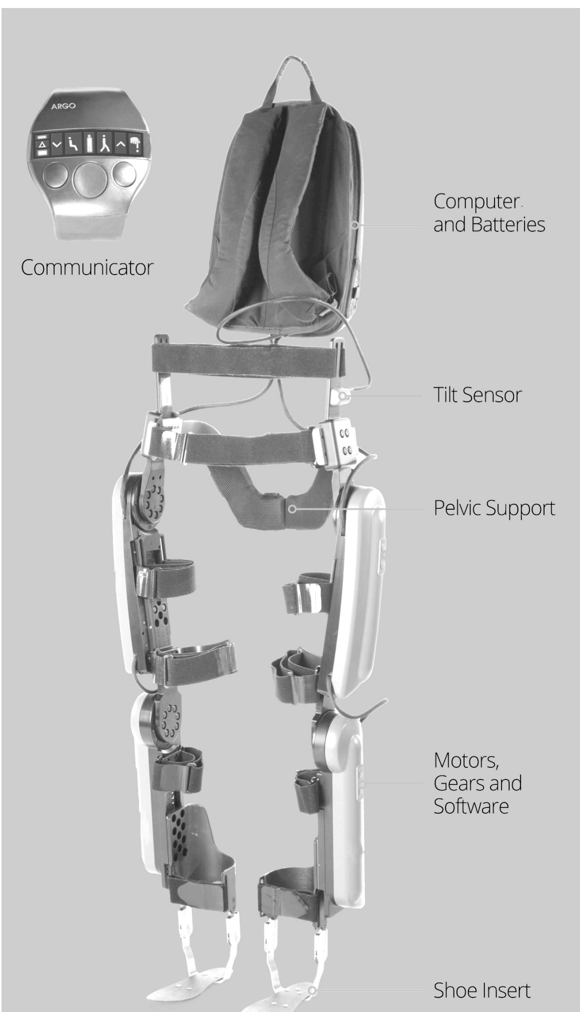

Fig. 1. The basic configuration of the ReWalk device.

II. Methods

A. Subject Selection

The research was approved by the IRB at the Albert Einstein Healthcare Network. Adults with chronic motor complete cervical and thoracic (C7-T12) spinal cord injury (according to American Spinal Injury Association (ASIA) guidelines) were recruited into the study. After an initial telephone screening, subjects came in to be consented for further assessment of their ability to participate. This consisted of Dual-energy X-ray absorptiometry (DXA), electrocardiography (ECG) and leg long bone and lumbar spine X-rays to confirm joint integrity and absence of unhealed fractures or heterotopic ossification that may impede walking. Subjects were required to have been standing (either with lower extremity bracing or a standing frame) on a frequent basis to be eligible. A complete neurological evaluation by a study co-PI was used to confirm injury level, skin integrity, hemodynamic stability, adequate hip, knee and ankle range of motion and a spasticity level of 3 or less using the Ashworth scale. The above criteria along with the absence of osteoporosis (basis of bone mineral density (BMD) > -2.5) at the right limb femoral neck and the L2 to L4 spine.

B. Basic Training with ReWalk^{TM}

The 12 subjects recruited into the study had never used the ReWalk^{TM} or any other exoskeleton before. They had no familiarity with the device. After being consented and passing all requisite medical criteria to participate, all began training learning to wear the device. Subjects were trained for up to 24 sessions of 60 to 90 minute duration over approximately 8 weeks (target was 3 times per week). Initial training consisted of learning to sit-to-stand, standing activities within parallel bars, stand-sit transfers, standing balance and stepping skills. Subsequently, training involved learning crutch use placement for balance and limb advancement. The remainder of the training aimed to improve and integrate walking performance with step triggering, coordinating step timing and foot clearance, and safe and effective stopping. Training was specific to each subject and followed their learning pace rather than a predetermined time table.

B. Data Collection and Analysis

Near the conclusion of the training, subjects were evaluated in the Gait & Motion Analysis Laboratory at MossRehab. There, they were instrumented using the active motion capture system (Coda CX1, Charnwood Dynamics, Ltd. England). Markers were placed in a modified protocol to account for the fact that anatomical regions were sometimes occluded by the ReWalk^{TM} exoskeleton. Data were collected as the subjects walked using their forearm crutches and the ReWalk^{TM} suit. Three-dimensional motion data, temporospatial data and dynamic EMG from selected lower extremity proximal muscles were collected during walking. Kinetic data were not obtained because of the need for crutches.

All subjects were able to independently walk, without human assistance while using the ReWalk^{TM}, for at least 50m continuously, for a period of at least 5-10 minutes continuously

ArgoMed tech partially funded the initial trial from which the data were generated and provided the suits for the study.

and with velocities ranging from 0.03m/s to 0.45m/s (mean 0.25m/s)[16]. After initial review of the data, subjects were divided into three groups, based on walking speed during the gait lab session.

Subsequently, data were processed following the laboratory clinical protocols and using the standard CODA Gait biomechanical model (CodaMA, v6.79, Charnwood Dynamics, Ltd. England). This enabled the calculation of angle-time histories for the trunk, pelvis, and bilateral hip, knee and ankle joints. These data were then exported from CodaMA to Microsoft Excel for further analysis. This analysis focused on the differences in above described body and joint angles with an emphasis on the time just prior to the start of swing phase and during swing phase as these were identified as the areas of likely critical differences to being a more successful ambulator.

III. RESULTS

The data obtained were stratified based on velocity using the following parameters : slow was between 0.22 and 0.31m/s (n=19, medium was between 0.32 and 0.41m/s (n=22) and fast was between 0.42 and 0.50m/s (n=4). Note that n refers to the number of trials, not subjects, in each group. The gait in ReWalk™ was fundamentally symmetric. Graphic data were generated for both sides (but not presented here for brevity). As such, we begin our analysis by looking at a single leg. Several key differences in segment and joint motion were observed across the three groups. The fast group showed a more flexed trunk at the start of swing phase (5-10 degrees over the medium and slow groups, Figure 2). In addition, the trunk flexion phasing appeared subtly different. In the fast group, peak flexion occurred at the start of swing phase where as in the other groups peak flexion occurred just before swing phase and extension had already started at swing initiation. A similar pattern was observed for pelvic tilt phasing but with opposite absolute positioning. The fast group showed a more extended (upright) pelvis throughout the entire gait cycle. Their overall profile oscillated around zero (neutral) with a slight bias towards extension. The other groups remained in flexion the entire gait cycle with an average value around 15 degrees. The timings were slightly more disparate than those seen in the trunk. The fast group showed peak anterior pelvic tilt just after swing phase began where as the other groups showed peak anterior pelvic tilt before or at the start of swing phase. It is noted that at ~15%, when the swing phase is just starting for the opposite leg, the trunk is considerably more flexed as well.

The fast group showed considerably greater hip extension over the entire gait cycle. In fact, they achieved full hip extension and even a bit more (~15 degrees of extension, on average) where as the slow group consistently showed full extension (into neutral position) but not more and the medium group only extended to 5 degrees of flexion (not even neutral position, Figure 3). At the start of swing phase, this difference was exacerbated. The fast group remained near peak extension (~10 degrees of extension, on average) where the other groups were closer to 10 degrees of flexion. Again, at ~15% (start of contralateral swing), the hip is more extended in the fast group than in the medium group (by about 10 degrees) and than in the slow group (by about 5 degrees).

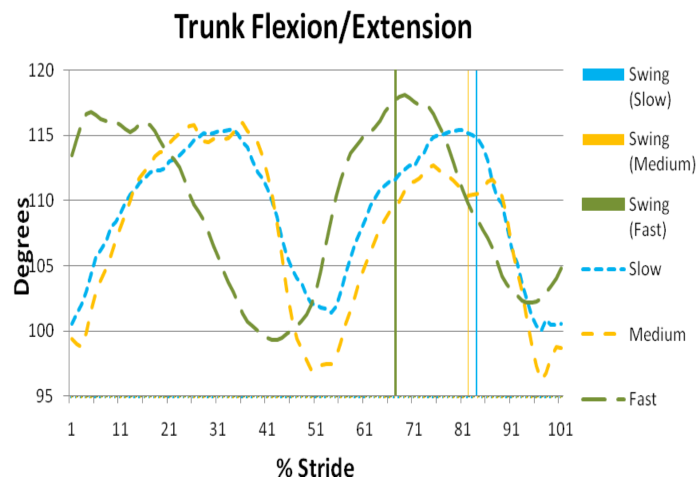

Fig. 2. Trunk flexion/extension for the three groups. Data were time normalized. The solid vertical lines demarcate the start of swing phase.

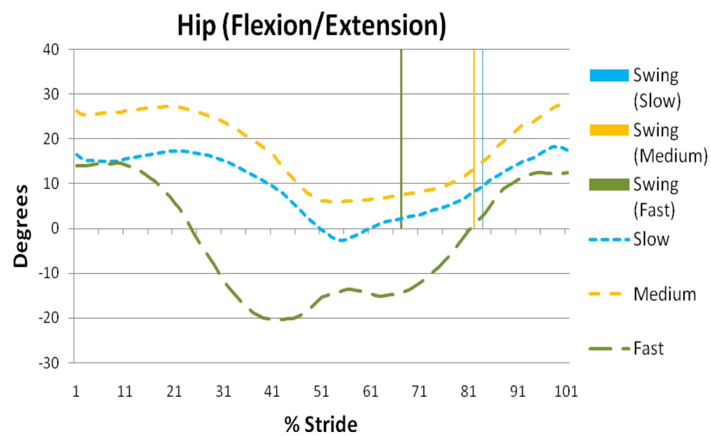

Fig. 3. Hip flexion/extension for the three groups. Data were time normalized. The solid vertical lines demarcate the start of swing phase.

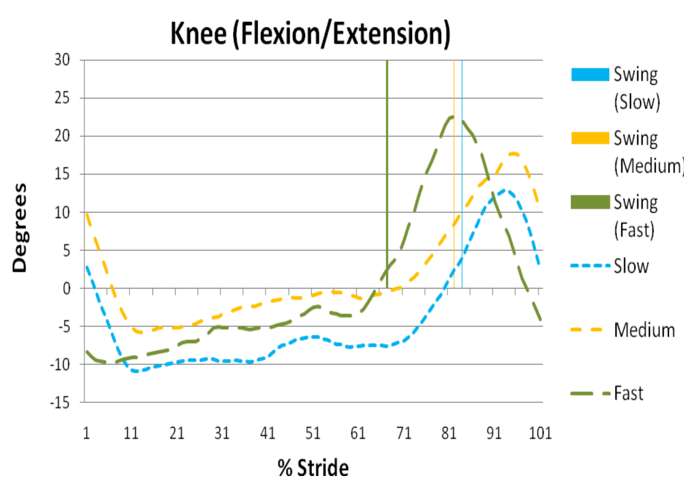

Fig. 4. Knee flexion/extension for the three groups. Data were time normalized. The solid vertical lines demarcate the start of swing phase.

All groups showed more similar patterns at the knee during stance phase, however the fast group was able to achieve considerably more flexion (~20 degrees) in swing phase compared to the medium (~15 degrees) and slow groups (~10 degrees, Figure 4). One other difference was that the fast group initiated swing with the knee more extended where as the other groups had already moved into flexion by a few degrees. At ~15% (start of contralateral swing), the knee is more extended in the fast and slow groups than in the medium group (by about 5 degrees).

At the ankle, the fast group showed increased plantarflexion in early stance phase, followed by steady dorsiflexion until just before swing phase and then a brief burst of plantarflexion. This 2nd plantarflexion peak occurred in swing phase and was also greater than that observed in the other groups. The medium and slow groups showed, coincidentally, about the same amount of plantarflexion at the start of swing with the key difference being they were actually dorsiflexing this ankle at this instant where as the fast group was plantarflexing. At ~15% (start of contralateral swing), the ankle is more extended (plantarflexed) in the fast group than in the slow and medium groups (by about 7-8 degrees).

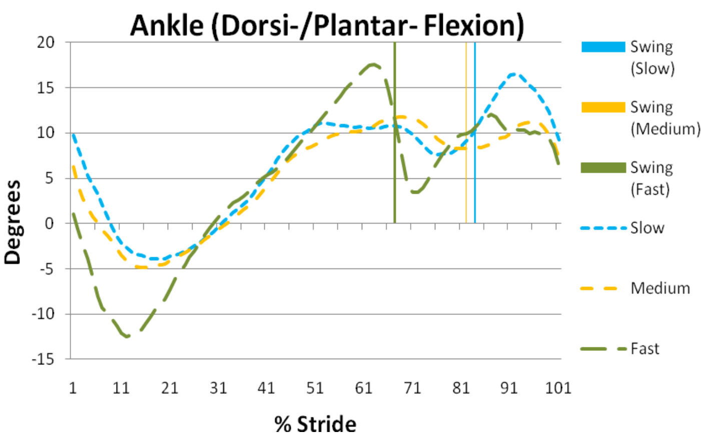

Fig. 5. Ankle plantarflexion and dorsiflexion for all three groups. Vertical lines demarcate the start of swing phase.

IV. DISCUSSION

Anecdotally, one major factor limiting the walking speed of the slower groups appeared to be their ability consistently achieve a good stepping pattern. Poor steps appeared to be caused by inability to consistently position the body to allow for good swing foot clearance and subsequent diminished step length due to premature foot contact with the ground. Understanding the differences between the faster and slower walkers may clarify how best to help the slower walkers to perform better. Thus, for the present analysis, we focused on the body positioning especially differences near the start of swing phase.

The gait in ReWalk™ was fundamentally symmetric. As such, we begin our analysis by looking at a single leg. At ~15% of the gait cycle in the graphs shown (Figs. 2-5) the contralateral leg (i.e. the leg not shown) would just be starting swing phase. So the positioning of the shown leg would contribute directly to the contralateral leg's ability to clear. We observed the hip and ankle of the stance leg to be more extended in the fast group – which would directly contribute to increased clearance of the swing leg. However, near the start of ipsilateral swing phase, we observed the stance leg to be more extended as well – which in normals would work against the ability to clear during the impending swing phase. Perhaps this is a reflection of the fast group's established confidence in obtaining clearance. Overall, there appear to be competing findings here – the generally increased extension at 15% of the early stance leg supports clearance but the increased extension observed in late stance would seem to counter clearance. At this point, we have not quantified these effects. Some fundamental kinematic calculations can clarify the contribution of these competing effects on actual clearance. In addition, the frontal plane kinematics can contribute to clearance in gait. Specifically, the pelvis can be raised to facilitate clearance – as is sometimes seen in gait pathologies with impaired clearance [17]. Furthermore, as is more likely the case in ReWalk™ subjects, whole body lateral leaning can be used to facilitate clearance. Lateral weight shifting was one aspect of early training. We are currently evaluating this aspect of the performance to see if and how it may have contributed to differences between groups.

The differences in walking speeds between the three groups were on the order of 0.1 and 0.2 m/s and appear relatively small – compared to normal walking speed for those without gait problems. However, the functional implications are significant. For example, the fast group is quite close to walking at the speed necessary to cross a busy urban street safely during the red light, where a 0.1m/s drop may not allow this. On the opposite end of the spectrum, the slowest ReWalkers could be limited in terms of how far they can walk, and ultimately in their functionality, by tiring due to inefficient gait. Furthermore, the slowest walkers may become frustrated and ultimately give up on walking with the ReWalk™. Though there was some loose correlation between level of injury and walking speed, much variability was observed in this relationship. It is our belief that some if not many of the slower walkers could learn to walk better (faster and with less energy expenditure). Doing so could open up the range and number of spinal cord injury patients who can be helped with this technology. Other causes of walking velocity differences need to be identified and their contribution explored as additional data becomes available.

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REFERENCES

[1] CDC. (2011, November 2, 2011). *Resources for Entertainment Education Content Developers*. Available: http://www.cdc.gov/healthcommunication/ToolsTemplates/EntertainmentEd/Tips/SpinalCordInjury.html

[2] M. Post, A. van Dijk, F. van Asbeck, and A. Schrijvers, "Life satisfaction of persons with spinal cord injury compared to a population group," *Scand J Rehab Med,* vol. 30, pp. 23-30, 1998.

[3] MJ Fuhrer, DH Rintala, KA Hart, R Clearman and ME Young, "Relationship of life satisfaction to impairment, disability, and handicap among persons with spinal cord injury living in the community," *Arch Phys Med Rehabil,* vol. 73, pp. 552-557, 1992.

[4] M. Dijkers, "Correlates of life satisfaction among persons with spinal cord injury," *Arch Phys Med Rehabil,* vol. 80, pp. 867-876, 1999.

[5] S. Kirshblum, M. Priebe, C. Ho, W. Scelza, A. Chiodo, and L. Wuermser, "Spinal cord injury medicine. 3. Rehabilitation phase after acute spinal cord injury," *Arch Phys Med Rehabil,* vol. 88, pp. S62-70, 2007.

[6] N. Welage and K. Liu, "Wheelchair accessibility of public buildings: a review of the literature," *Disabil Rehabil Assist Technol,* vol. 6, pp. 1-9, 2011.

[7] S. Goemaere, M. Van Laere, P. De Neve, and J. Kaufman, "Bone mineral status in paraplegic patients who do or do not perform standing.," *Osteoporos Int.,* vol. 4, pp. 138-43, 1994.

[8] J. Wall, "Preventing pressure sores among wheelchair users.," *Prof Nurse,* vol. 15, pp. 321-4, 2000.

[9] M. Regan, R. Teasell, D. Wolfe, D. Keast, W. Mortenson, and J. Aubut, "A systematic review of therapeutic interventions for pressure ulcers after spinal cord injury," *Arch Phys Med Rehabil,* vol. 90, pp. 213-31, 2009.

[10] S. Groah and I. Lanig, "Neuromusculoskeletal syndromes in wheelchair athletes," *Semin Neurol.,* vol. 20, pp. 201-8, 2000.

[11] M. Alm, H. Saraste, and C. Norrbrink, "Shoulder pain in persons with thoracic spinal cord injury: prevalence and characteristics.," *J Rehabil Med.,* vol. 40, pp. 277-83, 2008.

[12] R. Stein, S. Chong, K. James, A. Kido, G. Bell, L. Tubman, and M. Bélanger, "Electrical stimulation for therapy and mobility after spinal cord injury.," *Prog Brain Res.,* vol. 137, pp. 27-34, 2002.

[13] R. Chafetz, T. Johnston, and C. Calhoun, "Lower Limb Orthoses for Persons with Spinal Cord Injuriy," in *AAOS Atlas of Orthoses and Assistive Devices*, J. Hsu, J. Michael, and J. Fisk, Eds., 4th ed Philadelphia: Mosby, Elsevier Inc., 2008, p. chapter 27.

[14] L. Kaplan, B. Grynbaum, H. Rusk, T. Anastasia, and S. Gassier, "A reappraisal of braces and other mechanical aids in patients with spinal cord dysfunction: results of a follow-up study," *Arch Phys Med Rehabil,* vol. 47, pp. 393-405, 1966.

[15] M. Franceschini, Baratta, S, Zampolini, M, Loria, D, Lotta, S, "Reciprocating gait orthoses: a multicenter study of their use by spinal cord injured patients. Arch Phys Med Rehabil," vol. 78, pp. 582-586, Jun 1997.

[16] A. Esquenazi, M. Talaty, A. Packel, and M. Saulino, "The ReWalk Powered Exoskeleton to Restore Ambulatory Function to Individuals with Thoracic-Level Motor-Complete Spinal Cord Injury," *American Journal of Physical Medicine & Rehabilitation,* vol. 91, pp. 911-921, 2012.

[17] A. Esquenazi and M. Talaty, "Normal and Pathological Gait Analysis," in *Physical Medicine & Rehabilitation : The Complete Approach*, M. Grabois, S. J. Garrison, K. A. Hart, and L. D. Lehmkuhl, Eds., ed Malden, Massachusetts: Blackwell Science, Inc., 2000, p. 2002.