Simulator for Laser Photocoagulation in Ophthalmology

Patrick Dubois, Jean François Rouland, Philippe Meseure, Sylvain Karpf, and Christophe Chaillou

***Abstract*—The practice of laser photocoagulation plays a major role in the ocular therapy, but the persistence of many postoperative complications denotes genuine difficulty in mastering the technique. The authors present a device which, thanks to the use of simulation, enables actual practice to be dissociated from apprenticeship. While complying with the constraints of realism with regard to habitual conditions of laser use, the device offers access to a wide variety of clinical situations. The apparatus is built around the traditional instrument. A virtual image of the fundus is produced in real time from the sensors which detect the actual gestures used. The calculations make use of textured geometrical models. Digitized color photographs are organized to form a database which reflects the diversity of pigmentations and pathologies. A software interface has been developed to facilitate the use of the device. The prototype is operated using a PC-compatible computer; it displays the images at the rate of at least seven per second on a miniature CGA screen incorporated in the slit-lamp. It is currently being validated for clinical applications. Above and beyond apprenticeship in laser photocoagulation, its potential applications extend to the entire field of ophthalmogical symptomatology and, more broadly, to the simulation of any examination conducted with the help of binocular or endoscopic optics.**

I. Introduction

Since the first experiments on retinal photocoagulation, therapeutic indications using laser beams in medical and surgical pathology have extended considerably [1]–[10]. In the field of ocular therapy, laser plays a very important part (50 operations every week in the Regional University Hospital of Lille alone, for example). Now, despite the increase in the numbers of units installed in private practice, there has been no perceptible reduction in the proportion of retinal complications [1], [11], [12], resulting, in the worst cases, in complete blindness. Further operations then have to be envisaged, necessitating hospitalization for up to 4–5 days. This situation, in which heavy expense is involved, can be partly explained by failure to master the techniques.

Laser treatment requires a good command of the equipment and a perfect knowledge of incidences that could occur following its use. Under these circumstances, it is important to acquire the necessary skills in this field as the operator works unassisted and the risks of a manipulation error *in vivo* could have irreversible consequences. Unlike an audio-visual aid, a simulator, by taking account of gestures, modifies the scene observed almost instantaneously. It is a unique means, being entirely risk-free, of familiarizing future ophthalmologists with the conditions under which they will have to practice. Simulation further permits acquaintance with pathological situations that are both more diversified and more immediately accessible than under routine clinical conditions.

The advent of graphic modeling in medical imagery is quite recent [13]–[17] and the use of simulators in training doctors is still limited to endoscopy at the present time [18], [19]. The major aim of this work is to improve the training (both initial and ongoing) of ophthalmologists, which can only enhance the safety and efficiency of laser photocoagulation practiced in eye surgery using equipment that is not only fragile but requires fine adjustment and is expensive, and the handling of which necessitates special training.

Manuscript received August 1, 1994; revised March 20, 1995. This work was supported by DRED credits (EA 1049) and the allocation of research credits (Contract 9301) by the Regional and University Hospital of Lille (CH&U), as well as ANVAR Transfer Aid Contract A 93 11 121N.

P. Dubois is with CLARC-Laboratoire de Biophysique, Université de Lille II, France.

J. F. Rouland is with the Clinique Ophtalmologique, Hôpital Huriez, CH&U Lille, France.

P. Meseure, S. Karpf, and C. Chaillou are with the Laboratoire d'Informatique Fondamentale, Université de Lille I, France.

IEEE Log Number 9411747.

II. Material and Methods

A. Analysis and Modeling of Gestures

An instrument designed for training purposes has to comply with the usual conditions of operating practice. To minimize the effort of apprenticeship that has to be devoted to the simulator itself and to ensure optimum conformity with real operating conditions, the special features of the simulator have to be adapted to the equipment habitually used. In all aspects of this device, interdependence of the gestures made and visual observation plays an essential part; only observation of the location of the impact and the effect produced can guide ophthalmologists in their options.

The instruments used in laser photocoagulation comprise the following (Fig. 1):

1) A microscope through which the operator views the image of the fundus that is being treated; this sight is equipped with enlargement control optics.
2) A swingle bar permitting manual position control (right-left, up-down) of the sighting system in relation to the patient's eye. The front-rear translation movement adjusts the optical focus.
3) A device providing illumination by a slit-lamp producing a brightened height and width-adjustable rectangular area.
4) A laser source arriving at the surface to be coagulated via an optical path associated with the sighting system; a number of adjustments are possible (size, power), and it is triggered by a pedal.

0018-9294/95$04.00

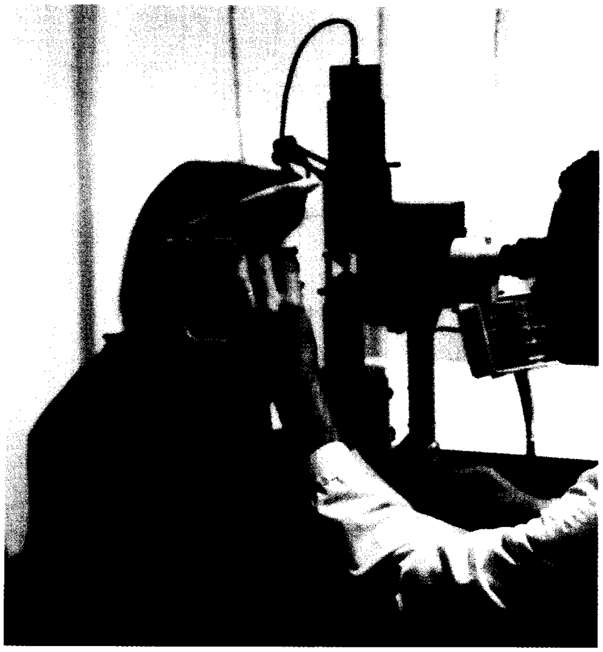


Fig. 1. Laser photocoagulator.

5) A micromanipulator used to control the deflection of the laser beam in relation to the optical axis of the device by two orthogonal rotations of a mirror.
6) A three-mirror lens (Fig. 2), which is the usual tool for clinical eye's investigation. It is manipulated by the operator and placed directly on the surface of the cornea. Its central lens (1) permits direct viewing of the posterior pole. The mirror (2) is inclined at 73° with regard to the frontal plane and permits the viewing of areas around the fundus. The mirror (3), inclined at 66°, permits the viewing of the peripheral areas. The mirror (4), inclined at 59°, permits the viewing of the areas around the iris. This three-mirror lens has three degrees of freedom, one rotation about its axis of revolution and the other two about the center of the cornea. Angular deflections about the eye are less than 27°. On the other hand, the lens rotates about its axis through 360°. The laser beam also reaches the retina through this lens.

The scene viewed by the practitioner depends on the gestures that are made using these different instruments. Some of these actions modify the geometry of the system (microscope translation, rotation of the examination lens, and deflection of the laser beam). Only the microscope and its swingle bar are physically present in our device, the other elements being simulated. A reference space had to be defined to describe the system, and we chose to associate it with the eye of the observer. It is in relation to this reference space that the center of the patient's eye, the center of rotation of the three-mirror lens (center of the cornea) and the center of deflection of the beam are identified. The left-right and up-down movements of the microscope are modeled by two orthogonal translation movements. These are described by a coordinate-changing matrix. The forward-reverse movement, acting on the optical focus, is modeled by an algorithm calculating the degree of focus which will be described in greater detail later. Optical enlargement is modeled by a zoom effect created by the software.

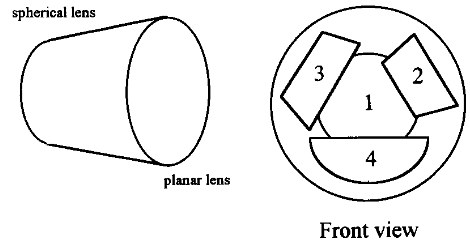


Fig. 2. The three-mirror lens.

The width and height adjustments of the rectangular-shaped area, illuminated by the split-lamp, are modeled by an algorithm which cuts a lighter colored size-varying rectangle in the total image of the retina.

The horizontal and vertical deflections of the laser spot are modeled by calculating the two rotations (φ and ψ) of the incident beam along the x and y axes, respectively. This makes it possible to calculate the point of impact on the deflection mirror, the possible reflection of this beam by one of the three mirrors and, finally, the intersection of this ray, direct or reflected, with the fundus. Manipulation of the three-mirror lens is described by two rotation matrices, the first rotation θ along the z-axis by M_θ

$$M_\theta = \begin{pmatrix} \cos\theta & -\sin\theta & 0 & 0 \\ \sin\theta & \cos\theta & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix}$$

the second, $M_{\varphi\psi}$

$$M_{\varphi\psi} = \begin{pmatrix} \cos\psi & \sin\varphi\sin\psi & \cos\varphi\sin\psi & 0 \\ 0 & \cos\varphi & -\sin\varphi & 0 \\ -\sin\psi & \sin\varphi\cos\psi & \cos\varphi\cos\psi & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix}$$

which reflects the combination of the other two rotations along the x and y axes. Each of the three mirrors of the lens is described by the equation of a plane.

B. Choice of Sensors

All the translation movements of the microscope are evaluated by linear potentiometer-type sensors integral mechanically with the observation statif. The rotations of the micromanipulator are measured using a potentiometer type joystick with two axes, fixed to the statif. The rotations of the three-mirror lens are measured using a potentiometer type joystick with three axes. A frustoconical sleeve is fixed to the spindle of an infinite rotation-type linear rotary potentiometer (Bourns$^{\text{TM}}$) which reflects the rotation along the z-axis. It is mounted on a unit with two double cradle-mounted universal joints, each associated with a rotary potentiometer. This assembly faithfully reflects the two rotations along the x and y axes, described in spherical coordinates according to the $M_{\varphi\psi}$ matrix. Control of the lighting slit, as well as of the

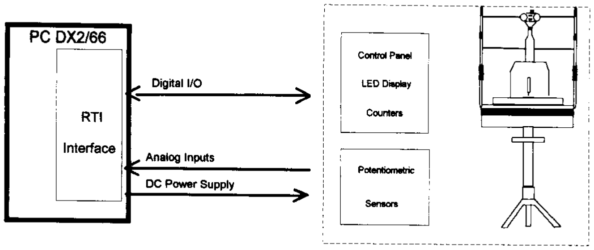


Fig. 3. Block diagram of the device.

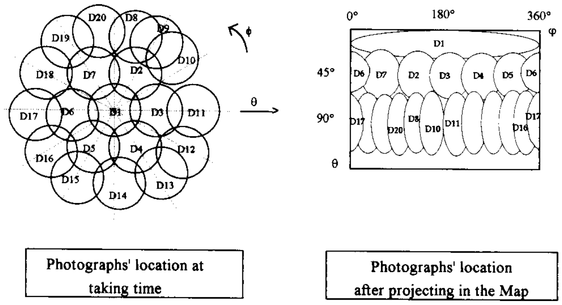


Fig. 4. Correspondence between photos and database.

diameter and luminous power of the laser beam, is converted by potentiometers. The variable dc voltages are fed into the computer via analog inputs of a conversion interface (RTI-835H Analog Device^{TM}), which encodes their values in 12-b format.

A six position selector switch, associated with encoding electronics, is connected to six binary inputs of the digital interface; it is used to determine the selected enlargement value. The simulated laser shot is triggered by a pedal connected to another binary input. This command is fed back to an events counter which serves to display (using LED's) the number of shots made, according to the standard practices.

All these sensors and their associated electronics are installed in a conventional examination device (Fig. 3), from which the slit-lamp and the laser tube have been removed. Two three-mirror lens joysticks are provided (left and right eyes) and inserted into the holes cut out for the eyes in a mask provided in place of the patient.

C. Simulation and Modeling of the Patient's Eye

In a first phase, an eye database (termed a "Map" at ϕ and θ) was formed from a set of 20 photographs for each fundus (transfer of photos to a Kodak Photo CD). A dedicated software was developed to fill in this "Map" (size: 1024×1024) by scanning all the dots on each photo, computing the coordinates of these dots in the image plane (projection of the sphere onto the plane), and copying the intensity value (R-G-B components, 12-b encoded). In its current version, this software is supervised by the operator to check, in particular, for continuity of vascular network and to verify any rotation of each photograph in relation to the common reference space (Fig. 4).

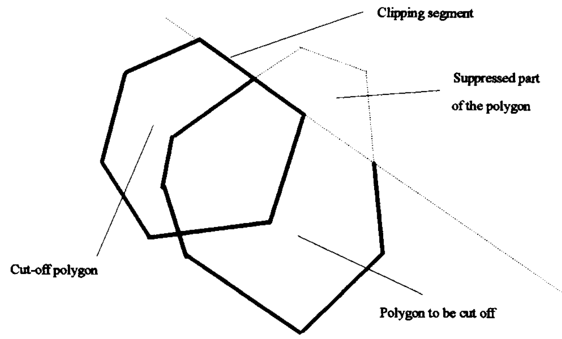


Fig. 5. Principle of the Sutherland-Hodgman algorithm.

In the second phase, assembly is carried out by calculating the changes in coordinates in the "Map" to image direction. It functions automatically. For focus control, three levels of focus texture are precomputed. The texture database is composed of four files in TGA format, sized $1024 \times 1024 \times 24$ b.

The eye is modeled by a 24-mm diameter sphere, the iris by a flat ring orthogonal to the optical axis, and the cornea by a hemisphere.

In order to ensure a computing speed compatible with real time operation, we divided up the eye sphere into facets, taking a regular icosahedron composed of 20 triangular facets, each facet then being textured. In the texture space, each triangle has its side and its height of 256 "texels," i.e., in all a table of $256^2 \times 20/2$ (655 360 bytes). A program is provided to obtain the textures associated with each triangle from the preceding TGA images.

D. Display

To avoid being confined to a particular machine, we chose to model each object simply and to display them in an unvarying order, making any sorting unnecessary: central lens and its edge, iris, fundus texture of the central lens, laser spot, mirrors, images of the central lens in the mirrors, images of the iris in the mirrors, images of the fundus in the mirrors, images of the laser spot in the mirrors, and trace of the reflection of the spot on the mirror. However, some objects have to be drawn solely inside other objects. For example, all the objects viewed through a mirror have to be cut off by the edges of this mirror. All the objects to be plotted are defined using convex polygons orientated in the trigonometric direction, delimiting the areas to be plotted. The algorithm used is that of Sutherland-Hodgman [20]; as in input, it requires orientated convex polygons and it produces in return a polygon that is also orientated—it can thus be used in series (Fig. 5).

Thus, the polygon of the pupil and that of the iris will be cut off by the polygon of the central lens. The polygon of the pupil obtained will then serve as a cut-off polygon for the fundus, which is seen only through the pupil.

The texture of the fundus is displayed by means of bilinear interpolation of the texel coordinates, which has the advantage of being very quick as it requires only two additions and two tests per pixel.

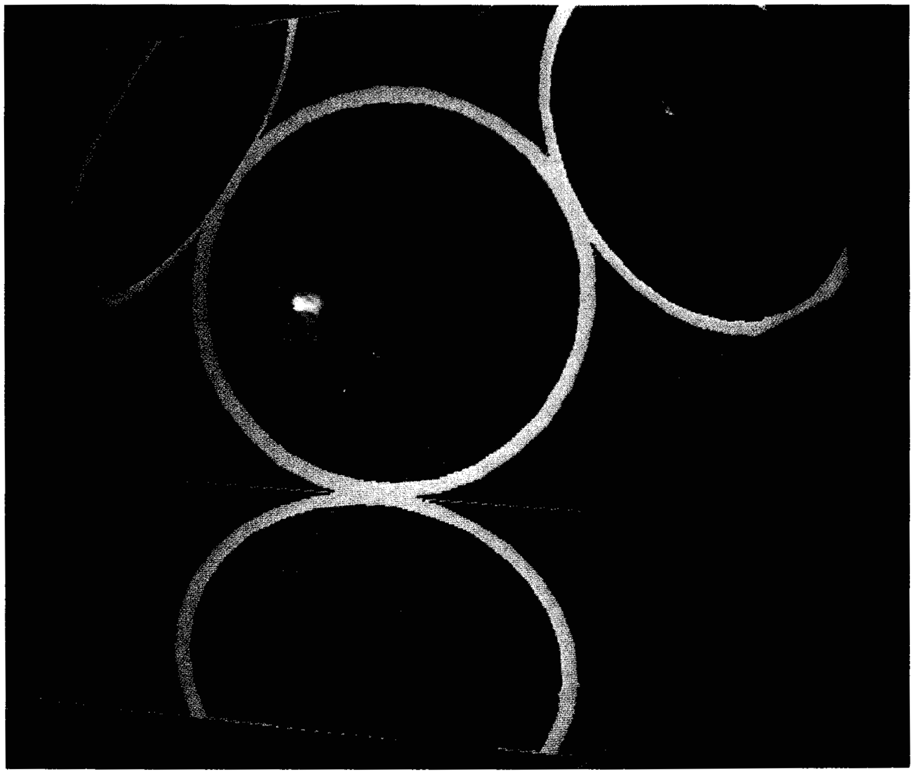


Fig. 6. Photograph of the monitor screen.

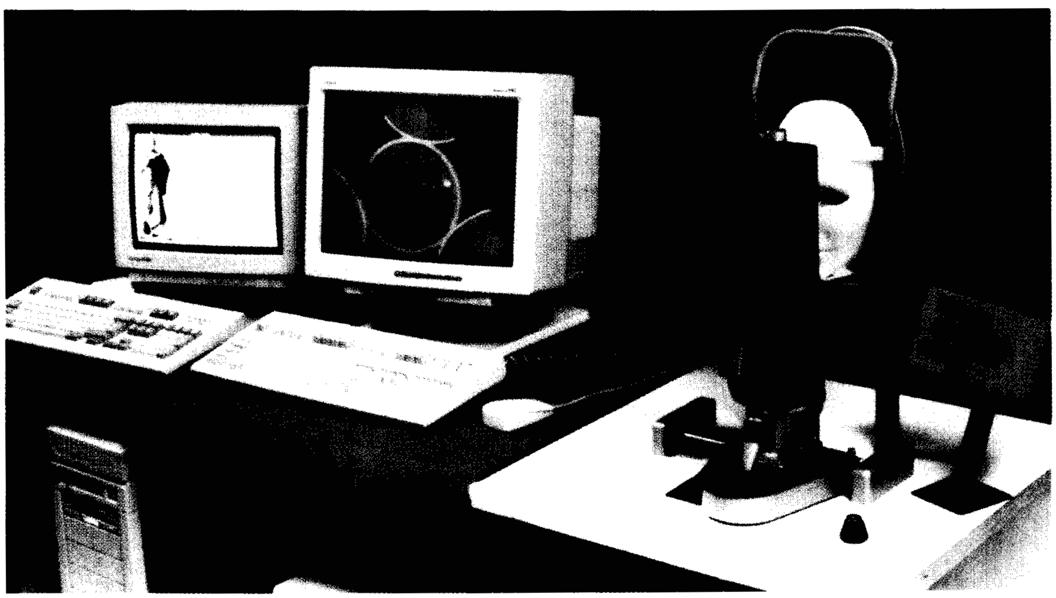


Fig. 7. The device.

To create the animation, a double buffer technique is used when plotting the image, that is, the visible screen and the plotted screen are different. The computing time depends on the size of the image to be plotted (enlargement, surface illuminated); it enables at least seven images per second to be obtained.

All of the objects to be plotted are illustrated by the photograph showing the monitor screen (Fig. 6).

E. User Interface

The simulator is used to provide photocoagulation sessions. These sessions are piloted by a succession of explicit menus designed to provide step by step assistance to users who are complete novices in data processing. The simulator is intended for two categories of user: the "experts" and the "students." The experts are the experienced practitioners to whom a number of students are allocated. The reference images are organized to create libraries: one library common to all those participating and one special library for each expert. This makes it possible to accommodate each "school of therapy" without imposing any standard.

These experts can either prepare a session for their students, choosing a fundus from their own libraries of pigmentations and pathologies, or prepare for themselves a delicate operation on a patient, the images for which they have previously recorded. A student, at beginner, intermediate, or advanced level, can select a session prepared for him/her by his/her expert or "replay" a previously completed session; each actual session is recorded, which also enables the expert to assess the work of the students off-line.

III. Design

The device (Fig. 7) is composed of two PC-compatible microcomputers linked together by a serial line.

1) The first is a type-486 DX2-66 model using MS-DOS, which pilots the operation of the simulator programs. It also handles the databases associated with the device. It is equipped with a Local Bus graphics card, the output of which is duplicated on two separate screens (a miniature screen incorporated in the microscope and a screen for monitoring by the expert). The images obtained have a resolution of 640×400, but the graphics resolution of the miniature screen (size: 6×14 mm^2) is still limited at present to the CGA standard. Following a 15-s initialization phase (for model calculation), the images are projected at a rate of at least seven per second.
2) The second computer provides the interface with the user via the teaching software. It runs under Windows and supervises the first computer, transferring to it the parameters specific to a particular performance of the session; it also receives the session recording images. Initial validation work has been undertaken with the cooperation of the interns in ophthalmology of the Lille Regional Hospital. A series of patents was filed in May 1994.

IV. Conclusions—Prospects

This simulator is initially used to acquire competence in retinal photocoagulation techniques. It could also be a teaching instrument for identification of fundus abnormalities liable to benefit from laser treatment. All pathologies amenable to laser photocoagulation can be simulated: [diabetic retinopathy, retinal vein occlusion, peripheral fundus abnormalities (lattice degeneration, retinal holes, and breaks or holes), and age-related macular degeneration].

All the parameters of photocoagulation (time, diameter, and power) are simulated as to their effects on the retina: overdosing, incorrect exposure time, malposition, diameter errors, and effects of healing of laser spots.

Furthermore, the simulator can project into the microscope all the ophthalmological symptoms: the eyelids, the anterior segment (conjunctiva, cornea, angle, lens, and iris), the posterior segment (vitreous body, retina, and optic nerve). In time, it will be possible to use this means to provide teaching on all ocular pathologies (corneal dystrophies, glaucomas, cataracts, and tumors). A number of ophthalmological centers of recognized competence in one of these fields will be contacted and asked to provide experts as well as to contribute to the creation of databases.

References

[1] J. Haut, S. Simon, M. Massin, and G. Perdriel, *Le Laser en Ophtalmologie.* Paris: Masson, 1981.

[2] F. A. L'Esperance, *Photocoagulation, Photoradiation and Surgery.* St. Louis, MO: Mosby, 1983.

[3] H. C. Zweng, "Laser in ophthalmology," in *Laser Applications in Medicine and Biology*, Wolbarsht, Ed. New York: Plenum, 1971, pp. 239–254.

[4] C. L. Schepens, *Retinal Detachment and Allied Diseases.* Philadelphia: Saunders, 1983.

[5] N. E. Byer, "Lattice degeneration of the retina," *Surv. Ophthalmol.*, vol. 23, pp. 213–248, 1979.

[6] ——, "The natural history of asymptomatic retinal breaks," *Ophthalmol.*, vol. 89, pp. 1033–1039, 1982.

[7] S. W. Hyams, E. Meir, M. Ivry, D. Krakowski, S. Barkai, E. Jedwab, and E. Neumann, "Chorioretinal lesions predisposing to retinal detachment," *Amer. J. Ophthalmol.*, vol. 78, pp. 429–437, 1974.

[8] G. Coscas, G. Soubrane, C. Ramahefasolo, and C. Fardeau, "Perifoveal laser treatment for subfoveal choroïdal new vessels in age-related macular degeneration. Results of a randomized clinical trial," *Arch. Ophthalmol.*, vol. 109, pp. 1258–1265, 1991.

[9] Macular Photocoagulation Study Group, "Five year follow-up of fellow eyes of patients with age-related macular degeneration and unilateral extrafoveal choroïdal neovascularisation," *Arch. Ophthalmol.*, vol. 111, pp. 1189–1199, 1991.

[10] ——, "Laser photocoagulation of subfoveal neovascular lesions of age-related macular degeneration: Updated findings from two clinical trials," *Arch. Ophthalmol.*, vol. 111, pp. 1200–1209, 1991.

[11] J. B. Mortimer, "The role of photocoagulation in retinal detachment," *Can. J. Ophthalmol.*, vol. 1, pp. 206–210, 1966.

[12] M. O. Yoshizumi, A. E. Kreiger, and D. M. Sharp, "Risk factors associated with the development of massive periretinal proliferation," *Retinal Diseases*, pp. 203–207, 1985.

[13] G. Wind, V. K. Dvorak, and J. A. Dvorak, "Computer graphic in modeling in surgery," *Orthop. Clin. North Amer.*, vol. 17, no. 4, pp. 657–668, 1986.

[14] C. N. McEwan and K. Fukuta, "Recent advances in medical imaging: Surgery planning and simulation," *World J. Surg.*, vol. 13, no. 4, pp. 343–348, 1989.

[15] N. M. Thalmann and D. Thalmann, "Toward virtual humans in medicine: A prospective view," *Comput. Med. Imag. Graph.*, vol. 18, no. 2, pp. 97–106, 1994.

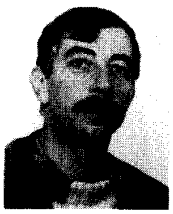
[16] S. L. Schwartz, Q. L. Cao, J. Azevedo, and N. G. Pandian, "Simulation of intraoperative visualization of cardiac structures and study of dynamic chirurgical anatomy with real-time tridimensional echocardiography," *Amer. J. Cardiol.*, vol. 73, no. 7, pp. 501–507, 1994.

[17] P. Clarysse *et al.*, "A computer-assisted system for localization in stereotaxic MRI," *IEEE Trans. Med. Imag.*, vol. 10, no. 4, pp. 523–529, 1991.

[18] M. Beergabel, S. Delmotte, and L. Muntlak, "Computer assisted training in endoscopy (CATE)—From a simulator to a training station," *Endoscopy*, vol. 24, suppl. 2, pp. 534–538, 1992.

[19] R. M. Satava, "Virtual reality surgical simulator. The first steps," *Surg. Endoscopy*, vol. 7, no. 3, pp. 203–205, 1993.

[20] J. Foley, A. Van Dam, S. Feiner, and J. Hughes, *Computer Graphics: Principle and Practice*, 2nd ed. Reading, MA: Addison-Wesley, 1990.

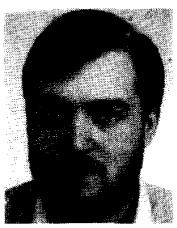


Patrick Dubois graduated as an Engineer from Institut Supérieur d'Electronique du Nord, Lille, in 1972. He received the Ph.D. in automatic control in 1974 from USTL (Université des Sciences et Technologies de Lille), France.

Since 1976, he has been an Assistant Professor at the Laboratoire de Biophysique, Faculté de Médecine, Lille and an Engineer in the Service Central de Médecine Nucléaire, Centre Hospitalier Régional, Lille. He is presently working at the Institut de Technologie Médicale, Centre Hospitalier Régional de Lille. His research and development interests are medical instrumentation and medical image analysis.

Dr. Dubois is a member of the IEEE Engineering in Medicine and Biology Society.

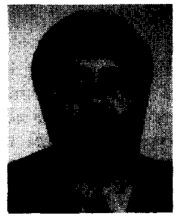
Jean François Rouland has been a Professor of Ophthalmology in the Huriez Hospital, University of Lille, France since 1991. His research activities are concerned with the glaucoma pathologies and the retinal photocoagulation.



Philippe Meseure received the M.Eng. in electronics at the Institut Supérieur d'Electronique du Nord, France, and the M.Sc. in computer science at LIFL, both in 1993. He is a Ph.D. student in computer graphics at the Laboratoire d'Informatique Fondamentale de Lille (LIFL), France. He is currently working on surgical, and especially ophthalmic, simulation. He is interested in physically based animation and real-time simulation.

Sylvain Karpf received the M.Sc. in computer science from USTL in 1989 and the Ph.D. in 1993.

He has been an Assistant Professor at the University of Lille I, France, since 1993. In the Computer Science Laboratory (LIFL), he is interested in 3-D real-time rendering, and more specifically, graphics hardware, VLSI design for real-time rendering, and medical simulation.



Christophe Chaillou received the Ph.D. in 1991 from USTL in computer science (massively parallel architectures for graphics hardware).

He has been an Assistant Professor at the University of Lille I, France, since 1991. His research interest in the Computer Science Laboratory (LIFL) is real-time visualization, including graphics hardware, medical simulators, and 3-D user interfaces for CSCW.