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(54) **FOCUSING ULTRASOUND INTO THE BRAIN THROUGH THE SKULL BY UTILIZING BOTH LONGITUDINAL AND SHEAR WAVES**

(75) Inventors: **Oleg Prus**, Haifa (IL); **Shuki Vitek**, Haifa (IL)

(73) Assignee: **Insightec Ltd.**, Tirat Carmel (IL)

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USPC **600/437**; 600/407; 600/442; 600/448

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USPC 310/328; 600/437–439, 441–443, 447, 600/449, 458, 459; 601/2, 3; 604/22; 73/620
See application file for complete search history.

(56) **References Cited**

U.S. PATENT DOCUMENTS

2,795,709	A		6/1957	Camp
3,142,035	A		7/1964	Harris
3,942,150	A		3/1976	Booth et al.
3,974,475	A		8/1976	Burckhardt et al.
3,992,693	A		11/1976	Martin et al.
4,000,493	A		12/1976	Spaulding et al.
4,074,564	A		2/1978	Anderson
4,206,653	A		6/1980	Lemay
4,221,132	A	*	9/1980	Poole 73/620
4,307,613	A		12/1981	Fox
4,339,952	A		7/1982	Foster
4,454,597	A		6/1984	Sullivan
4,478,083	A		10/1984	Hassler et al.
4,505,156	A		3/1985	Questo
4,526,168	A		7/1985	Hassler et al.
4,537,074	A		8/1985	Dietz
4,549,533	A		10/1985	Cain et al.
4,554,925	A		11/1985	Young
4,662,222	A		5/1987	Johnson
4,817,614	A	*	4/1989	Hassler et al. 600/441
4,858,597	A		8/1989	Kurtze et al.
4,865,042	A		9/1989	Umemura et al.
4,888,746	A		12/1989	Wurster et al.
4,889,122	A		12/1989	Watmough et al.

(Continued)

FOREIGN PATENT DOCUMENTS

DE	4345308 C2	2/2001
EP	1132054	9/2001

(Continued)

OTHER PUBLICATIONS

Clement et al., "A non-invasive method for focusing ultrasound through the human skull", Phys. Med. Biol. 47 (2002) pp. 1219-1236.*

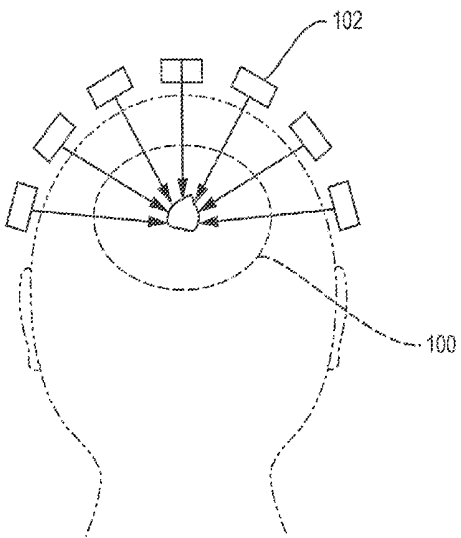
(Continued)

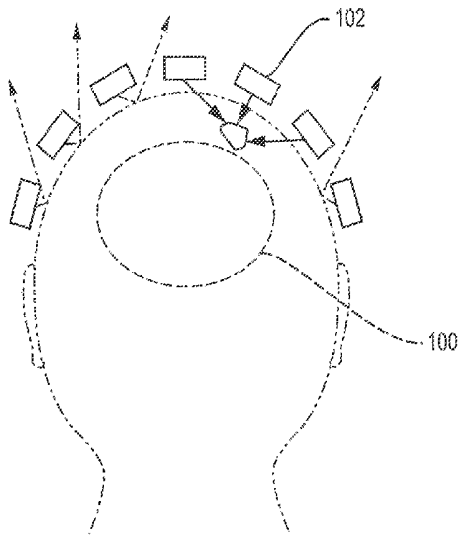
Primary Examiner — Unsu Jung
Assistant Examiner — Serkan Akar
(74) *Attorney, Agent, or Firm* — Bingham McCutchen LLP

(57) **ABSTRACT**

Systems and methods for focusing ultrasound through the skull into the brain for diagnostic or therapeutic purposes may be improved by utilizing both longitudinal and shear waves. The relative contribution of the two modes may be determined based on the angle of incidence.

23 Claims, 8 Drawing Sheets





(56) **References Cited**

U.S. PATENT DOCUMENTS

4,893,284 A 1/1990 Magrane
4,893,624 A 1/1990 Lele
4,937,767 A 6/1990 Reuschel et al.
5,197,475 A * 3/1993 Antich et al. 600/437
5,209,221 A 5/1993 Riedlinger
5,211,160 A 5/1993 Talish et al.
5,247,935 A 9/1993 Cline et al.
5,271,400 A 12/1993 Dumoulin et al.
5,275,165 A 1/1994 Ettinger et al.
5,291,890 A 3/1994 Cline et al.
5,307,812 A 5/1994 Hardy et al.
5,307,816 A 5/1994 Hashimoto et al.
5,318,025 A 6/1994 Dumoulin et al.
5,323,779 A 6/1994 Hardy et al.
5,327,884 A 7/1994 Hardy et al.
5,329,930 A 7/1994 Thomas, III et al.
5,368,031 A 11/1994 Cline et al.
5,368,032 A 11/1994 Cline et al.
5,379,642 A 1/1995 Reckwerdt et al.
5,391,140 A 2/1995 Schaetzle et al.
5,413,550 A 5/1995 Castel
5,435,312 A 7/1995 Spivey et al.
5,443,068 A 8/1995 Cline et al.
5,474,071 A 12/1995 Chapelon et al.
5,485,839 A 1/1996 Aida et al.
5,490,840 A 2/1996 Uzgiris et al.
5,507,790 A 4/1996 Weiss
5,520,188 A 5/1996 Hennige et al.
5,520,612 A * 5/1996 Winder et al. 601/2
5,526,814 A 6/1996 Cline et al.
5,549,638 A 8/1996 Burdette
5,553,618 A 9/1996 Suzuki et al.
5,573,497 A 11/1996 Chapelon
5,582,578 A 12/1996 Zhong et al.
5,590,653 A 1/1997 Aida et al.
5,590,657 A 1/1997 Cain et al.
5,601,526 A 2/1997 Chapelon et al.
5,605,154 A 2/1997 Ries et al.
5,606,971 A * 3/1997 Sarvazyan 600/438
5,617,371 A 4/1997 Williams
5,617,857 A 4/1997 Chader et al.
5,643,179 A 7/1997 Fujimoto
5,662,170 A 9/1997 Donovan et al.
5,665,054 A 9/1997 Dory
5,666,954 A 9/1997 Chapelon et al.
5,676,673 A 10/1997 Ferre et al.
5,687,729 A 11/1997 Schaetzle
5,694,936 A 12/1997 Fujimoto et al.
5,711,300 A 1/1998 Schneider et al.
5,722,411 A 3/1998 Suzuki et al.
5,728,062 A * 3/1998 Brisken 604/22
5,739,625 A 4/1998 Falcus
5,743,863 A 4/1998 Chapelon
5,752,515 A * 5/1998 Jolesz et al. 600/458
5,759,162 A 6/1998 Oppelt et al.
5,762,616 A 6/1998 Talish
5,769,790 A 6/1998 Watkins et al.
5,810,008 A 9/1998 Dekel et al.
5,810,731 A * 9/1998 Sarvazyan et al. 600/438
5,873,845 A 2/1999 Cline et al.
5,897,495 A 4/1999 Aida et al.
5,904,659 A * 5/1999 Duarte et al. 601/2
5,938,600 A 8/1999 Van Vaals et al.
5,938,608 A 8/1999 Bieger et al.
5,947,900 A 9/1999 Derbyshire et al.
5,984,881 A 11/1999 Ishibashi et al.
6,004,269 A 12/1999 Crowley et al.
6,023,636 A 2/2000 Wendt et al.
6,042,556 A 3/2000 Beach et al.
6,071,239 A 6/2000 Cribbs et al.
6,113,559 A 9/2000 Klopotek
6,128,522 A 10/2000 Acker et al.
6,128,958 A 10/2000 Cain
6,135,960 A * 10/2000 Holmberg 600/447
6,193,659 B1 2/2001 Ramamurthy et al.
6,217,530 B1 * 4/2001 Martin et al. 601/2
6,242,915 B1 6/2001 Hurd
6,246,896 B1 6/2001 Dumoulin et al.
6,263,230 B1 7/2001 Haynor et al.
6,267,734 B1 7/2001 Ishibashi et al.
6,289,233 B1 9/2001 Dumoulin et al.
6,309,355 B1 10/2001 Cain et al.
6,317,619 B1 11/2001 Boernert et al.
6,322,527 B1 11/2001 Talish
6,334,846 B1 1/2002 Ishibashi et al.
6,374,132 B1 4/2002 Acker et al.
6,392,330 B1 5/2002 Zloter et al.
6,397,094 B1 5/2002 Ludeke et al.
6,413,216 B1 7/2002 Cain et al.
6,419,648 B1 7/2002 Vitek et al.
6,424,597 B1 7/2002 Bolomey et al.
6,425,867 B1 7/2002 Vaezy et al.
6,428,532 B1 8/2002 Doukas et al.
6,433,464 B2 * 8/2002 Jones 310/328
6,461,314 B1 10/2002 Pant et al.
6,475,150 B2 11/2002 Haddad
6,478,739 B1 11/2002 Hong
6,506,154 B1 1/2003 Ezion et al.
6,506,171 B1 1/2003 Vitek et al.
6,511,064 B1 * 1/2003 Phinney et al. 271/262
6,511,428 B1 1/2003 Azuma et al.
6,522,142 B1 2/2003 Freundlich
6,523,272 B1 2/2003 Morales
6,524,251 B2 * 2/2003 Rabiner et al. 600/439
6,559,644 B2 5/2003 Froundlich et al.
6,566,878 B1 5/2003 Komura et al.
6,582,381 B1 6/2003 Yehezkeli et al.
6,599,256 B1 7/2003 Acker et al.
6,612,988 B2 9/2003 Maor et al.
6,613,004 B1 9/2003 Vitek et al.
6,613,005 B1 9/2003 Friedman et al.
6,618,608 B1 9/2003 Watkins et al.
6,618,620 B1 9/2003 Freundlich et al.
6,626,854 B2 9/2003 Friedman et al.
6,626,855 B1 9/2003 Weng et al.
6,629,929 B1 10/2003 Jago et al.
6,645,162 B2 11/2003 Friedman et al.
6,652,461 B1 11/2003 Levkovitz
6,666,833 B1 12/2003 Friedman et al.
6,676,601 B1 1/2004 Lacoste et al.
6,679,855 B2 1/2004 Horn et al.
6,705,994 B2 3/2004 Vortman et al.
6,719,694 B2 4/2004 Weng et al.
6,733,450 B1 5/2004 Alexandrov et al.
6,735,461 B2 5/2004 Vitek et al.
6,761,691 B2 7/2004 Tsuzuki
6,770,031 B2 8/2004 Hynynen et al.
6,770,039 B2 8/2004 Zhong et al.
6,788,619 B2 9/2004 Calvert
6,790,180 B2 9/2004 Vitek
6,824,516 B2 11/2004 Batten et al.
6,951,540 B2 10/2005 Ebbini et al.
6,961,606 B2 11/2005 DeSilets et al.
7,001,379 B2 2/2006 Behl et al.
7,077,820 B1 7/2006 Kadziauskas et al.
7,094,205 B2 8/2006 Marmarelis
7,128,711 B2 10/2006 Medan et al.
7,155,271 B2 12/2006 Halperin et al.
7,175,596 B2 2/2007 Vitek et al.
7,175,599 B2 * 2/2007 Hynynen et al. 600/443
7,264,592 B2 9/2007 Shehada
7,264,597 B2 9/2007 Cathignol
7,267,650 B2 9/2007 Chow et al.
7,344,509 B2 3/2008 Hynynen et al.
7,377,900 B2 5/2008 Vitek et al.
7,429,248 B1 * 9/2008 Winder et al. 601/2
7,452,357 B2 11/2008 Vlegele et al.
7,505,805 B2 3/2009 Kuroda
7,505,808 B2 3/2009 Anderson et al.
7,507,213 B2 * 3/2009 Schultheiss et al. 601/2
7,510,536 B2 3/2009 Foley et al.
7,511,501 B2 3/2009 Wexler
7,535,794 B2 5/2009 Prus et al.
7,553,284 B2 6/2009 Vaitekunas

(56) **References Cited**

U.S. PATENT DOCUMENTS

7,603,162	B2	10/2009	Danz et al.	
7,611,462	B2	11/2009	Vortman et al.	
7,652,410	B2	1/2010	Prus	
7,699,780	B2	4/2010	Vitek et al.	
2001/0031922	A1	10/2001	Weng et al.	
2002/0016557	A1*	2/2002	Duarte et al.	601/2
2002/0035779	A1	3/2002	Krieg et al.	
2002/0082589	A1	6/2002	Friedman et al.	
2002/0095087	A1*	7/2002	Mourad et al.	600/442
2002/0161300	A1*	10/2002	Hoff et al.	600/449
2002/0188229	A1	12/2002	Ryaby	
2003/0004439	A1	1/2003	Pant et al.	
2003/0060820	A1	3/2003	Maguire et al.	
2003/0187371	A1	10/2003	Vortman et al.	
2004/0030251	A1	2/2004	Ebbini et al.	
2004/0068186	A1	4/2004	Ishida et al.	
2004/0122316	A1	6/2004	Satoh	
2004/0122323	A1	6/2004	Vortman et al.	
2004/0143187	A1*	7/2004	Biagi et al.	600/437
2004/0210134	A1*	10/2004	Hynynen et al.	600/439
2004/0210135	A1*	10/2004	Hynynen et al.	600/439
2004/0236253	A1	11/2004	Vortman et al.	
2004/0267126	A1	12/2004	Takeuchi	
2005/0033201	A1	2/2005	Takahashi et al.	
2005/0096542	A1	5/2005	Weng et al.	
2005/0131301	A1	6/2005	Peszynski et al.	
2005/0203444	A1	9/2005	Schonenberger et al.	
2005/0251046	A1	11/2005	Yamamoto et al.	
2006/0052661	A1	3/2006	Gannot et al.	
2006/0052701	A1	3/2006	Carter et al.	
2006/0058678	A1	3/2006	Vitek et al.	
2006/0106300	A1	5/2006	Seppenwoolde et al.	
2006/0173385	A1	8/2006	Lidgren et al.	
2006/0184034	A1*	8/2006	Haim et al.	600/459
2006/0184069	A1*	8/2006	Vaitekunas	601/2
2006/0206105	A1	9/2006	Chopra et al.	
2006/0229594	A1	10/2006	Francischelli et al.	
2007/0016039	A1	1/2007	Vortman et al.	
2007/0055140	A1	3/2007	Kuroda	
2007/0066897	A1	3/2007	Sekins et al.	
2007/0073135	A1	3/2007	Lee et al.	
2007/0098232	A1	5/2007	Matula et al.	
2007/0167781	A1	7/2007	Vortman et al.	
2007/0197918	A1	8/2007	Vitek et al.	
2007/0219470	A1	9/2007	Talish et al.	
2007/0276237	A1	11/2007	Li	
2008/0027342	A1	1/2008	Rouw et al.	
2008/0031090	A1*	2/2008	Prus et al.	367/13
2008/0033278	A1	2/2008	Assif	
2008/0082026	A1	4/2008	Schmidt et al.	
2008/0108900	A1	5/2008	Lee et al.	
2008/0183077	A1	7/2008	Moreau-Gobard et al.	
2008/0228081	A1	9/2008	Becker et al.	
2008/0312562	A1	12/2008	Routh et al.	
2009/0088623	A1	4/2009	Vortman et al.	
2009/0118619	A1	5/2009	Oshiki	
2010/0030076	A1	2/2010	Vortman et al.	
2010/0056962	A1	3/2010	Vortman et al.	
2010/0125193	A1	5/2010	Zadicario	
2010/0179425	A1	7/2010	Zadicario	
2010/0268088	A1*	10/2010	Prus et al.	600/459
2010/0318002	A1	12/2010	Prus et al.	

FOREIGN PATENT DOCUMENTS

EP	1591073		11/2005
EP	1774920	A1	4/2007
EP	1790384		5/2007
EP	1936404		6/2008
FR	2806611	A1	9/2001
JP	5-92008		4/1993
JP	7-184907		7/1995
JP	7-231895		9/1995
JP	7-313518		12/1995
JP	11313833	A	11/1999
JP	00/166940		6/2000
JP	01/516075		9/2001
JP	02/530145		9/2002
WO	WO-9100059	A1	1/1991
WO	WO-9852465	A1	11/1998
WO	WO-00/031614		6/2000
WO	WO-0031614	A1	6/2000
WO	WO-0166189	A1	9/2001
WO	WO-01/080709		11/2001
WO	WO-0180709	A2	11/2001
WO	WO-02/43805		6/2002
WO	WO-02058791	A1	8/2002
WO	WO-03/013654		2/2003
WO	WO-03/013654	A1	2/2003
WO	WO-03097162	A2	11/2003
WO	WO-03098232	A2	11/2003
WO	WO-2004/093686		11/2004
WO	WO-200558029	A2	6/2005
WO	WO-2006/018837		2/2006
WO	WO-2006018837	A2	2/2006
WO	WO-2006025001	A1	3/2006
WO	WO-2006087649	A1	8/2006
WO	WO-2006/119572		11/2006
WO	WO-2007073551	A1	6/2007
WO	WO-2008/039449		4/2008
WO	WO-2008050278	A1	5/2008
WO	WO-200875203	A2	6/2008
WO	WO-2008119054	A1	10/2008
WO	WO-2009055587	A1	4/2009
WO	WO-2009/094554		7/2009
WO	WO-2010/058292		5/2010
WO	WO-2010/082135		7/2010
WO	WO-2010/119340		10/2010
WO	WO-2010/143072		12/2010
WO	WO-2011/013001		2/2011
WO	WO-2011/024074		3/2011

OTHER PUBLICATIONS

Clement et al., "Enhanced ultrasound transmission through the human skull using shear mode conversion" J. Acoust. Soc. Am. 115 (3), Mar. 2004, pp. 1356-1364.*

McGough et al., "Direct Computation of Ultrasound Phased-Array Driving Signals from a Specified Temperature Distribution for Hyperthermia," IEEE Transactions on Biomedical Engineering, vol. 39, No. 8, pp. 825-835 (Aug. 1992).

Botros et al., "A hybrid computational model for ultrasound phased-array heating in presence of strongly scattering obstacles," IEEE Trans. on Biomed. Eng., vol. 44, No. 11, pp. 1039-1050 (Nov. 1997).

Cain et al., "Concentric-ring and Sector-vortex Phased-array Applicators for Ultrasound Hperthermia," IEEE Trans. on Microwave Theory & Techniques, vol. MTT-34, No. 5, pp. 542-551 (May 1986).

Chen et al., "MR Acoustic Radiation Force Imaging: Comparison of Encoding Gradients.", (2008).

Cline et al., "Focused US system for MR imaging-guide tumor ablation," Radiology, v. 194, No. 3, pp. 731-738 (Mar. 1995).

Cline et al., "MR Temperature mapping of focused ultrasound surgery," Magnetic Resonance in Medicine, vol. 32, No. 6, pp. 628-636 (1994).

Cline et al., "Simultaneous magnetic resonance phase and magnitude temperature maps in muscle," Magnetic Resonance in Medicine, vol. 35, No. 3, pp. 309-315 (Mar. 1996).

Daum et al., "Design and evaluation of a feedback based phased array system for ultrasound surgery," IEEE Trans. Ultrason. Ferroelec. Freq. Control, vol. 45, No. 2, pp. 431-434 (1998).

de Senneville et al., "Real-time adaptive methods for treatment of mobile organs by MRI-controlled high-intensity focussed Ultrasound," Magnetic Resonance in Medicine 57:319-330 (2007).

Fjield et al, "The Combined Concentric-ring and Sector-vortex Phased Array for MRI Guided Ultrasound Surgery," IEEE Trans. on Ultrasonics, Ferroelectrics and Freq. Cont., vol. 44, No. 5, pp. 1157-1167 (Sep. 1997).

Herbert et al., "Energy-based adaptive focusing of waves: application to ultrasonic transcranial therapy," 8th Intl. Symp. on Therapeutic Ultrasound, (2009).

(56) **References Cited**

OTHER PUBLICATIONS

Huber et al., "A New Noninvasive Approach in Breast Cancer Therapy Using Magnetic Resonance Imaging-Guided Focussed Ultrasound Surgery," Cancer Research 61, 8441-8447 (Dec. 2001).
International Preliminary Report on Patentability in International Patent Application No. PCT/IB2004/001512, mailed Dec. 8, 2005.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2004/001498, dated Aug. 31, 2004.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2005/002273, mailed Dec. 20, 2005.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2005/002413, mailed Nov. 22, 2005.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2006/001641, mailed Sep. 25, 2006.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2006/003300, mailed Feb. 14, 2008.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2007/001079, mailed Dec. 10, 2007.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2007/002134, mailed Dec. 13, 2007.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2007/002140, mailed Dec. 29, 2008.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2008/003069, mailed Apr. 27, 2009.
Jolesz et al., "Integration of interventional MRI with computer-assisted surgery," J. Magnetic Resonance Imaging. 12:69-77 (2001).
Kohler et al., "Volumetric HIFU Ablation guided by multiplane MRI thermometry," 8th Intl. Symp. on Therapeutic Ultrasound, edited by E.S. Ebbini, U. of Minn. (Sep. 2009).
Kowalski et al., "Optimization of electromagnetic phased-arrays for hyperthermia via magnetic resonance temperature estimation," IEEE Trans. on Biomed. Eng., vol. 49, No. 11, pp. 1229-1241 (Nov. 2002).
Maxwell et al., "Noninvasive thrombolysis using pulsed ultrasound cavitation therapy—Histotripsy," Abstract, U.S. Natl. Lib. of Med., NIH, Ultrasound Med. Biol. (Oct. 23, 2009).
McDannold et al., "MRI evaluation of thermal ablation of tumors and focused ultrasounds," JMRI vol. 8, No. 1, pp. 91-100 (1998).
McDannold et al., "Magnetic resonance acoustic radiation force imaging," Med. Phys. vol. 35, No. 8, pp. 3748-3758 (Aug. 2008).
Medel et al., "Sonothrombolysis: An emerging mordality for the management of stroke," Neurosurgery, vol. 65, No. 5, pp. 979-993, (2009).
Mougenot et al., "MR monitoring of the near-field HIFU heating," 8th Intl. Symp. on Therapeutic Ultrasound, edited by E.S. Ebbini, U. of Minn. (Sep. 2009).
Partial International Search Report and Written Opinion in International Patent Application No. PCT/IB2007/001079, dated Sep. 25, 2007.
Vimeux et al., "Real-time control of focused ultrasound heating based on rapid MR thermometry," Investig. Radiology, vol. 43, No. 3, pp. 190-193, (1999).
Vykhodtseva et al., "MRI detection of the thermal effects of focused ultrasound on the brain," Ultrasound in Med. & Biol., vol. 26, No. 5, pp. 871-880 (2000).
Written Opinion in International Patent Application No. PCT/IL01/00340, mailed Feb. 24, 2003.
Written Opinion in International Patent Application No. PCT/IL02/00477, mailed Feb. 25, 2003.
Written Opinion in International Patent Application No. PCT/IB03/05551, mailed Sep. 10, 2004.
"How is Ablatherm treatment performed?" http://www.edap-hifu.com/eng/physicians/hifu/3c_treatment_treat-description.htm, accessed Jan. 3, 2003.
"What is HIFU? HIFU: High Intensity Focused Ultrasound," http://www.edap-hifu.com/eng/physicians/hifu2a_hifu_overview.htm, accessed Jan. 3, 2003.
"What are the physical principles?" http://www.edap-hifu.com/eng/physicians/hifu/2c_hifu_physical.htm, accessed Jan. 3, 2003.
"How does HIFU create a lesion?" http://www.edap-hifu.com/eng/physicians/hifu/2d_hifu_lesion.htm, accessed Jan. 3, 2003.
"Prostate Cancer Phase I Clinical Trials Using High Intensity Focused Ultrasound (HIFU)," Focus Surgery, http://www.focus-surgery.com/PCT%20Treatment%20with%20HIFU.htm, accessed Jan. 3, 2003.
"Abstract" Focus Surgery, http://www.focus-surgery.com/Sanghvi.htm, accessed Jan. 3, 2003.
Exablate 2000 Specification, InSightec, Ltd. (2 pages), (2002).
FDA Approves Exablate 2000 as Non-invasive surgery for Fibroids, Oct. 22, 2004.
International Search Report and Written Opinion in International Patent Application No. PCT/IB2010/000189, mailed Jun. 1, 2010.
International Search Report for PCT/IB03/05551 completion date Mar. 2, 2004 (5 pages).
International Search Report and Written Opinion in Internation Patent Application No. PCT/IB2010/000971, mailed Jul. 29, 2010 (9 pages).

* cited by examiner

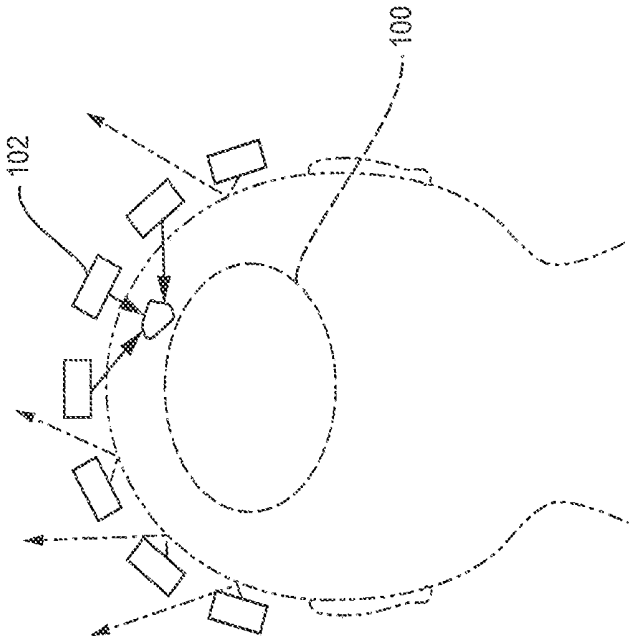


FIG. 1B

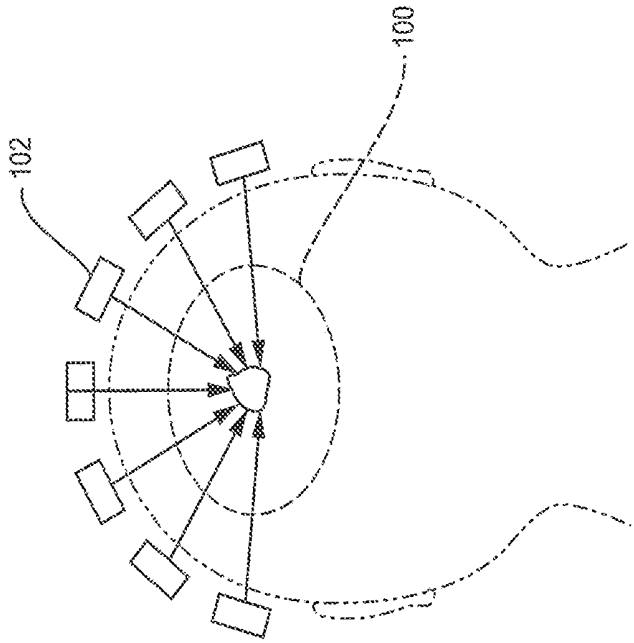


FIG. 1A

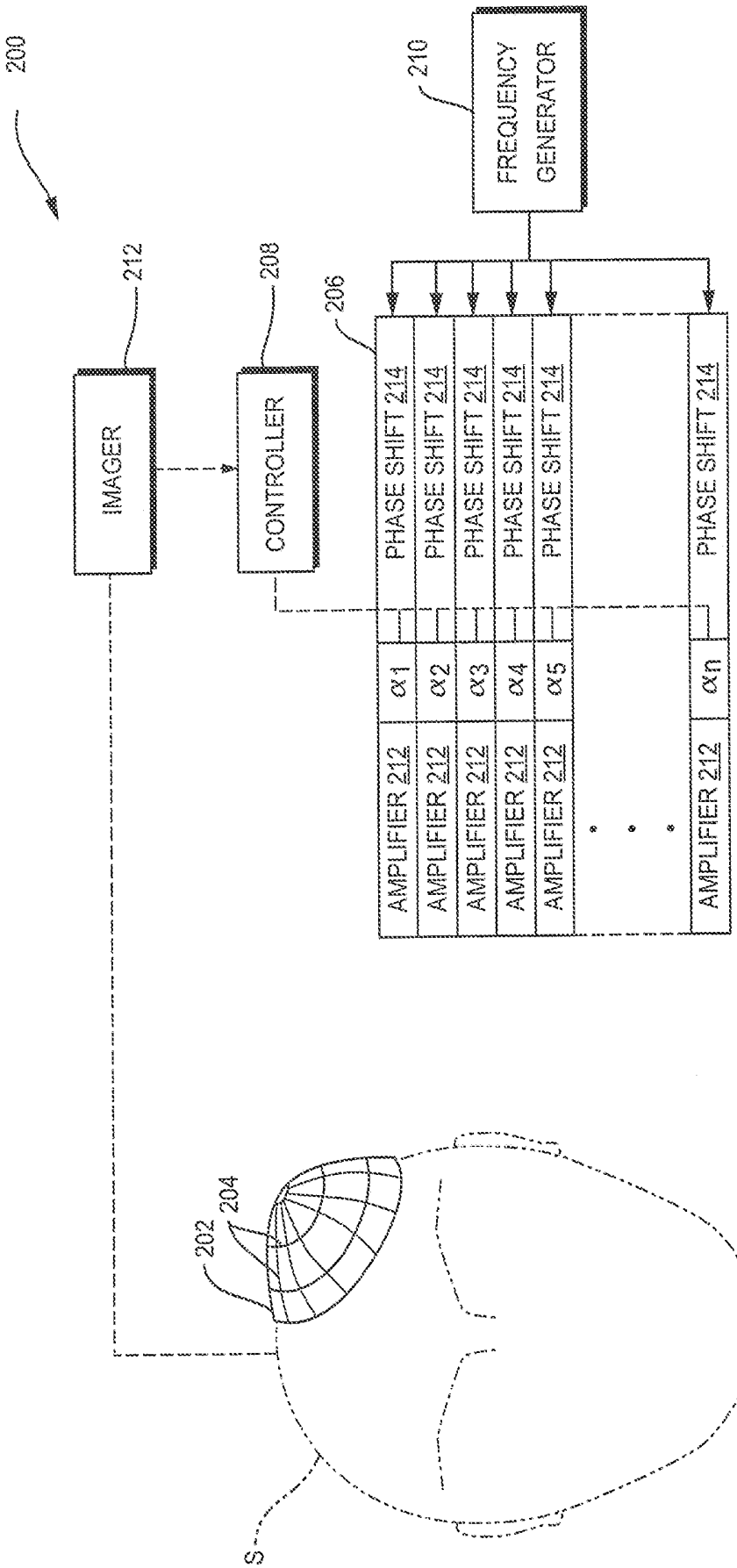

200
210
FREQUENCY GENERATOR
212
208
206
IMAGER
CONTROLLER
PHASE SHIFT 214
AMPLIFIER 212
α1
α2
α3
α4
α5
αn
202
204
S

FIG. 2

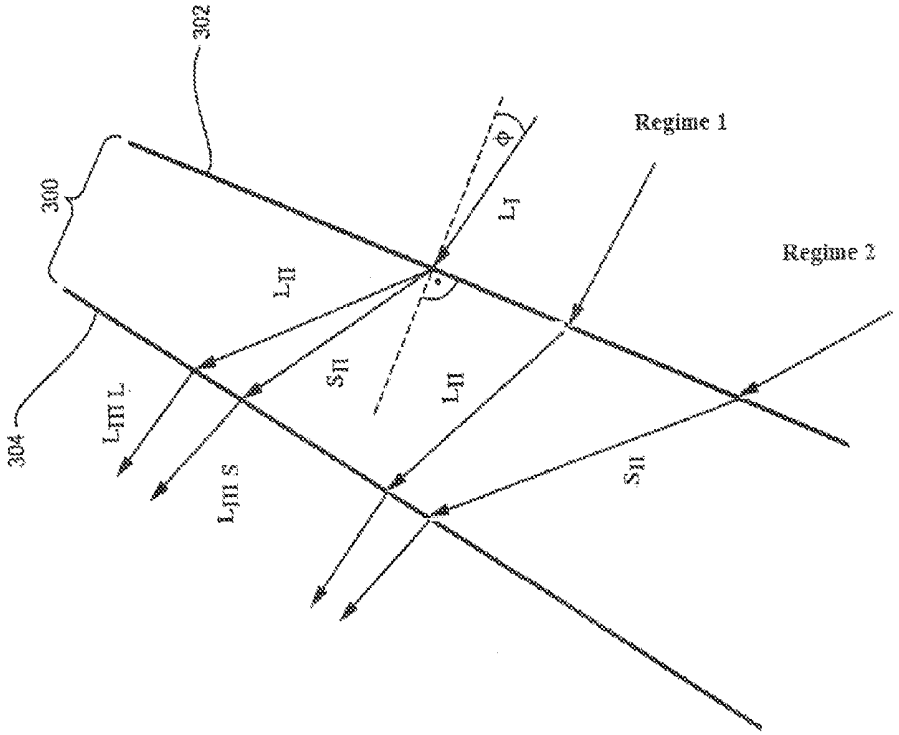

302
300
304
Regime 1
Regime 2
Φ
L_I
L_{II}
S_{II}
$L_{III\ L}$
$L_{III\ S}$

FIG. 3B

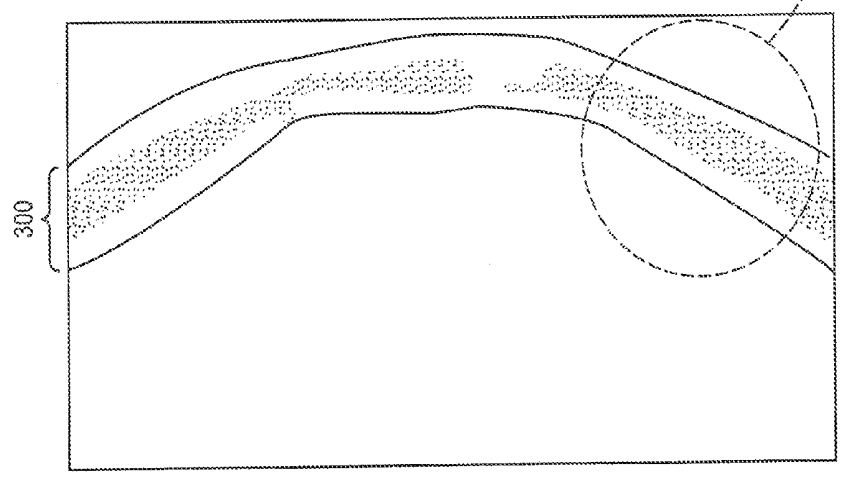


FIG. 3B
300
300

FIG. 3A

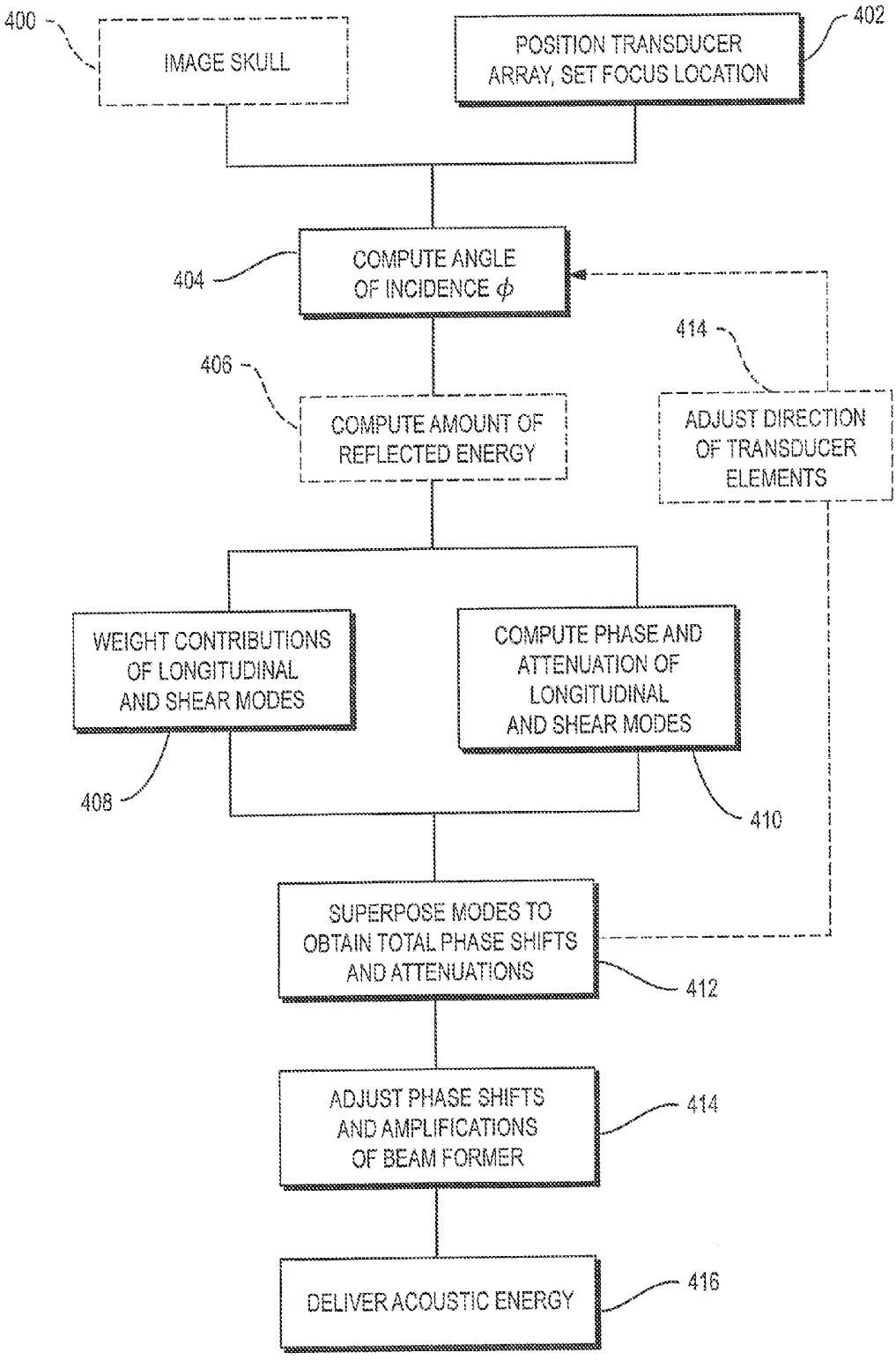

400
IMAGE SKULL
402
POSITION TRANSDUCER ARRAY, SET FOCUS LOCATION
404
COMPUTE ANGLE OF INCIDENCE ϕ
414
406
COMPUTE AMOUNT OF REFLECTED ENERGY
ADJUST DIRECTION OF TRANSDUCER ELEMENTS
WEIGHT CONTRIBUTIONS OF LONGITUDINAL AND SHEAR MODES
COMPUTE PHASE AND ATTENUATION OF LONGITUDINAL AND SHEAR MODES
408
410
SUPERPOSE MODES TO OBTAIN TOTAL PHASE SHIFTS AND ATTENUATIONS
412
ADJUST PHASE SHIFTS AND AMPLIFICATIONS OF BEAM FORMER
414
DELIVER ACOUSTIC ENERGY
416

FIG. 4

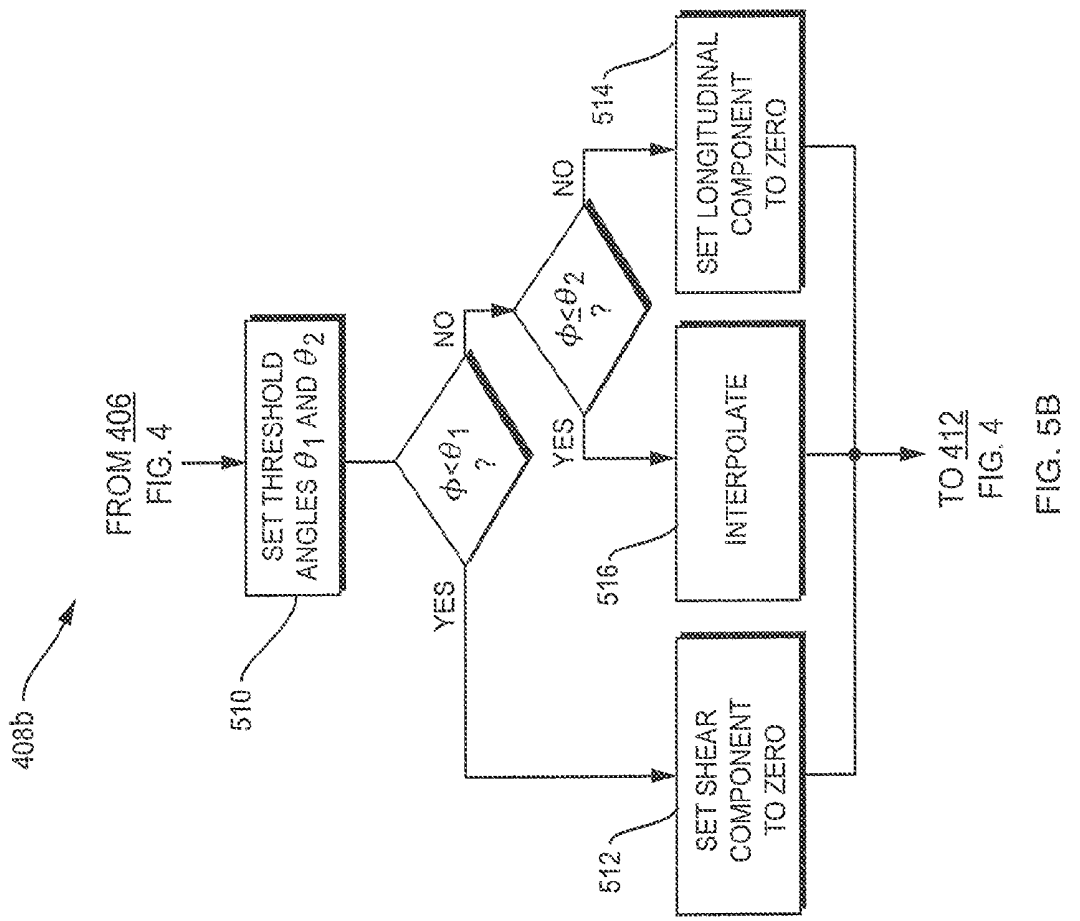

408b
FROM 406
FIG. 4
SET THRESHOLD ANGLES θ_1 AND θ_2
510
$\phi<\theta_1$?
YES
NO
$\phi\leq\theta_2$?
YES
NO
512
SET SHEAR COMPONENT TO ZERO
516
INTERPOLATE
514
SET LONGITUDINAL COMPONENT TO ZERO
TO 412
FIG. 4

FIG. 5B

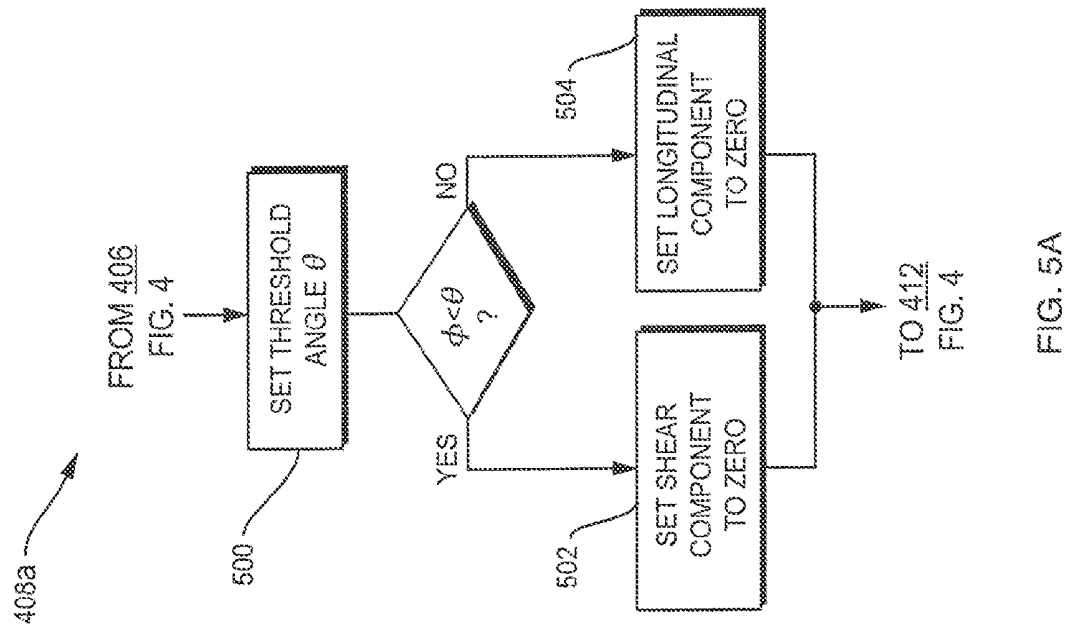

408a
FROM 406
FIG. 4
SET THRESHOLD ANGLE θ
500
$\phi<\theta$?
YES
NO
502
SET SHEAR COMPONENT TO ZERO
504
SET LONGITUDINAL COMPONENT TO ZERO
TO 412
FIG. 4

FIG. 5A

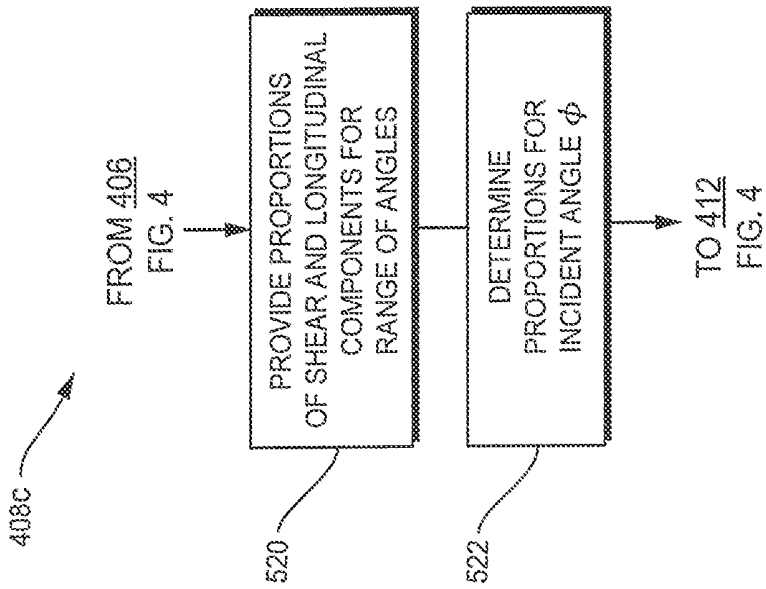

FROM 406
FIG. 4
PROVIDE PROPORTIONS OF SHEAR AND LONGITUDINAL COMPONENTS FOR RANGE OF ANGLES
DETERMINE PROPORTIONS FOR INCIDENT ANGLE ϕ
TO 412
FIG. 4
408C
520
522

FIG. 5C

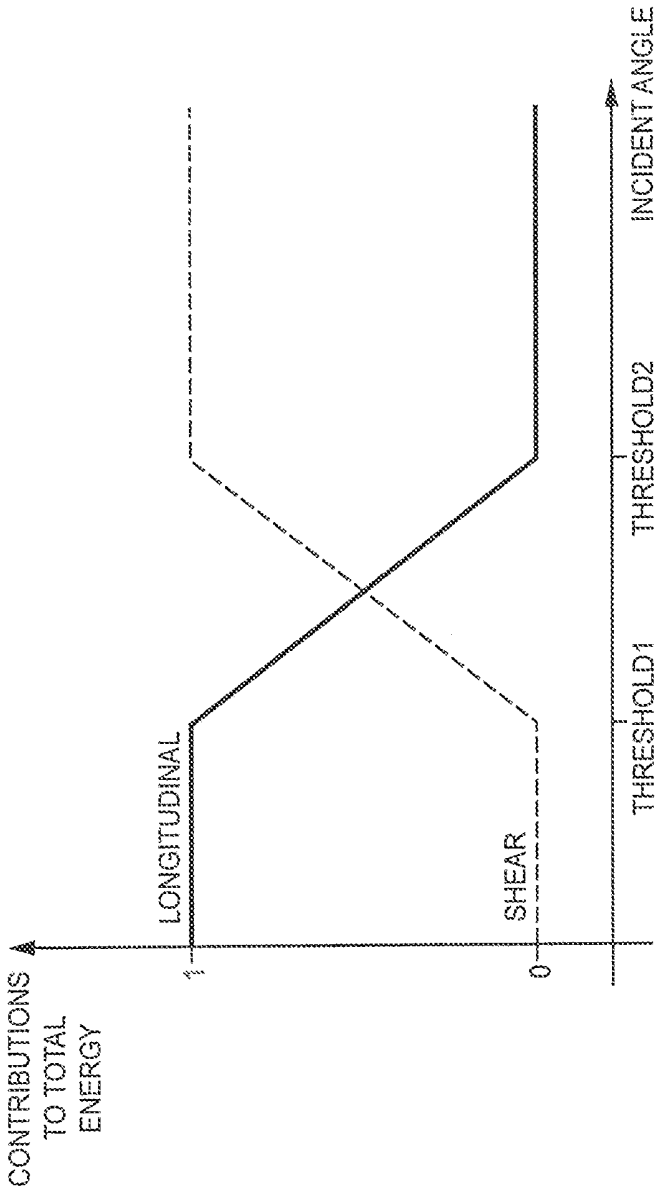

CONTRIBUTIONS TO TOTAL ENERGY
1
0
LONGITUDINAL
SHEAR
THRESHOLD1
e.g. 20°
THRESHOLD2
e.g. 40°
INCIDENT ANGLE

FIG. 6

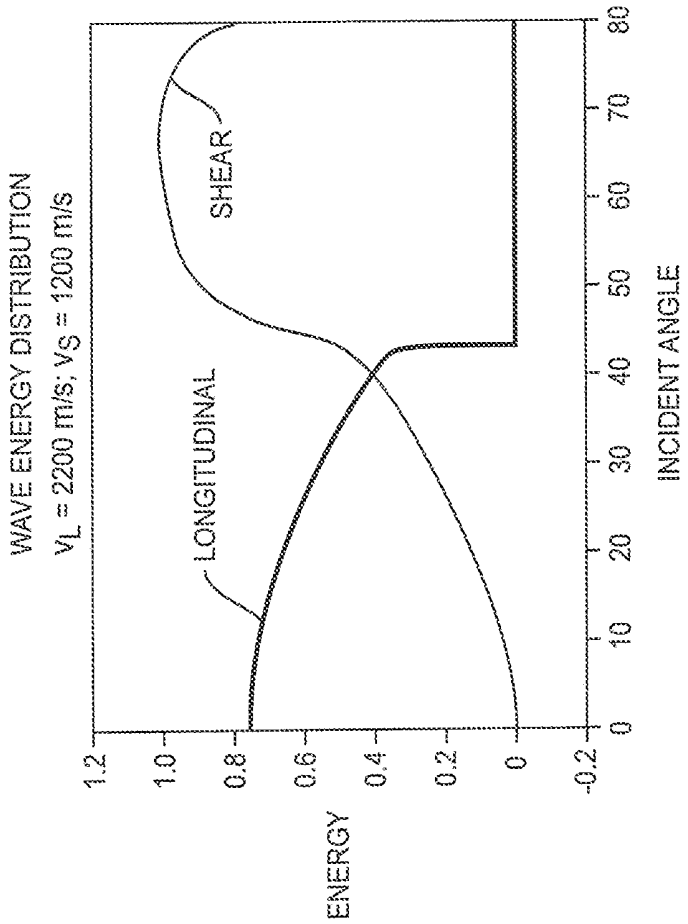

WAVE ENERGY DISTRIBUTION
v_L = 2200 m/s; v_S = 1200 m/s
LONGITUDINAL
SHEAR
ENERGY
INCIDENT ANGLE
1.2
1.0
0.8
0.6
0.4
0.2
0
-0.2
0
10
20
30
40
50
60
70
80

FIG. 7

FOCUSING ULTRASOUND INTO THE BRAIN THROUGH THE SKULL BY UTILIZING BOTH LONGITUDINAL AND SHEAR WAVES

BACKGROUND

Ultrasound penetrates well through soft tissues and, due to its short wavelengths, can be focused to spots with dimensions of a few millimeters. As a consequence of these properties, ultrasound can and has been used for a variety of diagnostic and therapeutic medical purposes, including ultrasound imaging and non-invasive surgery of many parts of the body. For example, by heating diseased (e.g., cancerous) tissue using ultrasound, it is often possible to ablate the diseased portions without causing significant damage to surrounding healthy tissue.

The noninvasive nature of ultrasound surgery is particularly appealing for the treatment of brain tumors. Moreover, coherent, non-invasive focusing of ultrasound through the human skull has been considered as a tool for targeted drug delivery to the brain, improved thrombolytic stroke treatment, blood flow imaging, the detection of internal bleeding, and tomographic brain imaging. However, the human skull has been a barrier to the clinical realization of many of these applications. Impediments to transcranial procedures include strong attenuation and the distortions caused by irregularities in the skull's shape, density, and sound speed, which contribute toward destroying the ultrasound focus and/or decreasing the ability to spatially register received diagnostic information.

Several minimally invasive or noninvasive aberration-correction techniques for transskull focusing overcome the focusing difficulties at least partially. Minimally invasive approaches may use receiving probes designed for catheter insertion into the brain to measure the amplitude and phase distortion caused by the skull, and then correct the ultrasound beam using an array of transducers. An alternative, completely noninvasive approach uses X-ray computed tomography (CT) images, rather than receiving probes, to predict the wave distortion caused by the skull.

Noninvasive focusing with a therapeutic array has been demonstrated at frequencies of about 2 MHz with a longitudinal wave propagation model. The velocity of these waves is approximately 2700 m/s in the skull, and about 1500 m/s in water and soft tissue. Due to this ratio, sound that arrives at the skull under an incident angle above about 30°, the critical angle, is reflected. The amplitude of the focus therefore drops when the focus is directed close to the skull surface. As shown in FIGS. **1**A and **1**B, a treatment envelope **100** is defined as the region accessible to ultrasound from a sufficient number of transducers **102** to enable treatment. Whereas the effect of reflection is minimal when the target area is deep within the brain (FIG. **1**A), since it lies inside the treatment envelope **100**, reflection becomes problematic when the target is outside the envelope **100** (FIG. **1**B). In the latter case, only a small number of transducers can reach the target area without reflection, and the closer the target is to the skull, the more transducers will be completely excluded from treatment due to reflection.

The treatment envelope **100** can be extended by reducing the frequency, e.g., to 0.2 MHz, and employing shear waves. Shear waves are largely absorbed in the skull at frequencies between 0.5 MHz and 4 MHz; at lower frequencies, however, their absorption is reduced to about that of longitudinal modes. Moreover, at 0.2 MHz, the sound velocity of shear waves in water (~1500 m/s) is comparable to that in the skull (~1400 m/s), thereby essentially eliminating the problem of reflection above a critical angle.

Previous methods of utilizing shear waves have calculated the phase shifts and amplitude attenuation associated with an originally longitudinal mode that is converted to a shear mode upon encountering the skull, and converted back to a longitudinal mode when entering the soft tissue of the brain. This approach is limited to large incidence angles at which no longitudinal mode is excited in the skull, or is otherwise inaccurate. In order to optimize focusing properties and maximize the amount of energy available in the focus, the coexistence of longitudinal and transverse modes ought to be taken into consideration.

SUMMARY

The present invention provides, in various embodiments, methods and apparatus for ultrasound focusing utilizing both longitudinal and shear modes. While developed mostly for non-invasive brain surgery and imaging, the approach of the invention may also be applied to other parts of the body requiring the penetration of ultrasound through a bone or cartilage interface. In transcranial applications, a transducer array emits longitudinal sound waves, which split into longitudinal and transverse components when entering the skull. The ratio of these two components depends on the frequency and the incidence angle. At the interface of the skull with the soft brain tissue, the transverse wave, or shear wave, is converted back to a longitudinal wave, or pressure wave, and both longitudinal components contribute to the ultrasound focus in the brain. Methods embodying the invention compute and correct for a phase shift and/or amplitude attenuation experienced by the waves during their propagation through the skull and brain.

In a first aspect, the invention provides, in various embodiments, a method for delivering ultrasound waves to a target through at least one tissue layer, which defines a first and a second interface. The waves have both a shear component and a longitudinal component through the first tissue layer. The method involves weighting energy contributions from the shear component and the longitudinal component based on an incident angle of the waves at the first interface and, optionally, further based on a frequency of the waves and/or on an acoustic response of the first layer. The method may also include the computation of a coefficient for the reflection of the waves at the first interface of the first tissue layer. Furthermore, the method includes the steps of computing phase shifts, and optionally attenuation coefficients, associated with the shear component and the longitudinal component, and compensating for the phase shifts and/or attenuations based on the weightings when delivering the ultrasound waves. The compensation step may, in some embodiments, also account for the geometry, thickness, density, and/or acoustic response of the first tissue layer. This first tissue layer corresponds in certain embodiments to a patient's skull. The method may involve providing a model of the first tissue layer (and any additional tissue layers), on which the compensation step may be based. Such a model may be obtained using computer tomography or magnetic resonance imaging. In certain embodiments, the model contains locally parallel layers.

In some embodiments, the weighting of energy contributions from the shear and longitudinal components is accomplished by determining a threshold angle, and setting the contribution from the shear component to zero for incident angles below the threshold angle and the contribution from the longitudinal component to zero for incident angles above or at the threshold angle. Alternatively, two threshold angles

may be determined. In these alternative embodiments, the contribution from the shear mode is set to zero for incident angles below the smaller of the two threshold angles, and the contribution of the longitudinal mode is set to zero for angles above the larger of the two threshold angles. For incident angles between the threshold angles, the energy contributions of the two modes are computed by (e.g., linear) interpolation. In some embodiments, the weighting step is based on a relationship between energy contributions from the shear component and the longitudinal component that has been determined empirically over a range of incidence angles.

In certain embodiments, the ultrasonic wave incident upon the first interface is purely longitudinal, and excites a longitudinal and a shear component in the first tissue layer. The shear component, in turn, may excite a longitudinal wave at the second interface.

In various embodiments, the ultrasound waves are delivered by a phased array of transducers, whose outputs are adjusted according to the computed phase shifts and, if applicable, attenuations in the compensation step. The waves may have a frequency in the range from 100 kHz to 400 kHz.

In a second aspect, the invention provides, in various embodiments, a system for delivering multimode ultrasound waves to a target located inside an object that contains at least a first tissue layer defining a first and a second interface. The system includes a phased array of transducers arranged around the object, a controller computing phase shift compensations for the transducers, and a beam former in communication with the phased array of transducers for adjusting outputs of the transducers in accordance with the phase shift compensations computed by the controller. The controller receives data about the object and its location relative to the transducers, as well as about the location of the target, and uses this data to compute an incident angle at which waves emanating from the transducer arrive at the first interface. Based on the computed incidence angle, the controller computes weightings of the energy contributions from the shear component and the longitudinal component. It further calculates phase shifts for the two wave components, based on the incident angle and the data about the object, and uses these phase shifts in determining the phase shift compensations.

BRIEF DESCRIPTION OF THE DRAWINGS

The foregoing discussion and the following detailed description of embodiments of the invention can more readily be understood in conjunction with the accompanying drawings wherein:

FIGS. **1**A and **1**B are schematic diagrams illustrating ultrasound wave focusing for foci in the center and near the periphery of the brain, respectively;

FIG. **2** is a schematic diagram of a system for focusing ultrasound in accordance with various embodiments of the invention;

FIG. **3**A is a computer tomography image of a skull section, and FIG. **3**B is a schematic diagram of an enlarged view of the skull section illustrating acoustic wave propagation through the skull;

FIG. **4** is a flow chart illustrating a method for ultrasound focusing longitudinal and transverse modes in accordance with various embodiments of the invention;

FIGS. **5**A-**5**C are flow charts detailing the weighting step of FIG. **4** in accordance with various embodiments of the invention;

FIG. **6** is a graph illustrating an interpolation for the energy distribution between longitudinal and transverse waves in accordance with one embodiment of the invention; and

FIG. **7** is a graph showing empirical data for the energy distribution between longitudinal and transverse waves which may be used in accordance with one embodiment of the invention.

DETAILED DESCRIPTION

Ultrasound energy may be transmitted into a patient, for various medical purposes, using arrays of radiating transducer elements. The amplitudes of the waves emanating from the elements can affect how much energy penetrates the patient, and the relative phases and amplitudes of the waves can be controlled to focus the energy into a desired region, e.g., into a tumor. While penetrating the patient, the waves typically encounter several layers of tissues, e.g., bone, muscle, or fat, whose density and structure, and, consequently, ultrasound propagation properties, differ. Due to inhomogeneities and anisotropies in the tissues, the wave fronts are often distorted. Moreover, signals from different transducer elements may encounter different thicknesses and contours of materials, and possibly air-filled or liquid-filled pockets between transducer elements and the region to be imaged or treated, resulting in different phase shifts and attenuations. Compensating for these effects, by appropriate phase shifts and amplification factors imposed on the transducer elements, avoids deterioration of focusing properties.

FIG. **2** illustrates an exemplary ultrasound therapy system **200** for focusing ultrasound into a patient's brain through the skull S. The system **200** includes a phased array **202** of transducer elements **204**, a beam former **206** driving the phased array, a controller **208** in communication with the beamformer, and a frequency generator **210** providing an input electronic signal for the beamformer. The system may further include an imager **212**, such as a magnetic resonance imaging (MRI) device or a computer tomography (CT) device, for determining the structure of the patient's skull and brain.

The array **202** may comprise a single row or a matrix of transducer elements **204**. In alternative embodiments, the transducer elements **204** may be arranged in a non-coordinated fashion. The array **202** may have a curved (e.g., spherical or parabolic) shape suitable for placing it on the surface of the skull S, or may include one or more planar or otherwise shaped sections. Its dimensions may vary, depending on the application, between millimeters and tens of centimeters. The transducer elements **204** of the array **202** may be piezoelectric ceramic elements, and may be mounted in silicone rubber or any other material suitable for damping the mechanical coupling between the elements **204**. Piezo-composite materials, or generally any materials capable of converting electrical energy to acoustic energy, may also be used. To assure maximum power transfer to the transducer elements **204**, the elements may be configured for electrical resonance at 50Ω, matching input connector impedance.

The array **202** is coupled to the beamformer **206**, which drives the individual transducer elements so that they collectively produce a focused ultrasonic beam or field. For n transducer elements, the beamformer **206** may contain n pairs of an amplifier **212** and a phase delay circuit **214**, each pair driving one of the transducer elements. The beamformer **206** receives a radio frequency (RF) input signal, typically in the range from 0.1 MHz to 0.4 MHz, from frequency generator **210**, which may, for example, be a Model DS345 generator available from Stanford Research Systems. The input signal may be split into n channels for the n amplifiers and delay circuits **212**, **214** of the beamformer **206**. In some embodiments, the frequency generator **210** is integrated with the

beamformer **206**. The radio frequency generator **210** and the beamformer **206** are configured to drive the individual transducer elements **204** of the array **206** at the same frequency, but at different phases and different amplitudes.

The amplification or attenuation factors and the phase shift α imposed by the beamformer **206** serve to transmit and focus ultrasonic energy through the patient's skull S into a selected region of the patient's brain, and account for wave distortions induced in the skull and soft brain tissue. They are computed in the controller **208**, which may provide the computational functions through software, hardware, firmware, hardwiring, or any combination thereof. For example, the controller **208** may utilize a general purpose, or special purpose, digital data processor programmed with software in a conventional manner in order to determine the phase shifts and amplification factors. In certain embodiments, the computation is based on detailed information about the structure, thickness, density, etc. of the skull. Such information may be obtained from the imager **212**. Image acquisition may be three-dimensional or, alternatively, the imager **212** may provide a set of two-dimensional images suitable for constructing a three-dimensional image of the skull and brain from which thicknesses and densities can be inferred. Image-manipulation functionality may be implemented in the imager **212**, in the controller **208**, or in a separate device.

System **200** may be modified in many ways within the scope of the invention. For example, for diagnostic applications, the system may further include a detector device that measures transmitted or reflected ultrasound, and which may provide the signals it receives to the driver **208** for further processing. The reflection and transmission signals may also be used as feedback for the phase and amplitude adjustments of the beamformer **206**. The system may contain a positioner for arresting the array **202** of transducer elements **204** with respect to the patient's skull S. In order to apply ultrasound therapy to body parts other than the brain, the transducer array **202** may take a different, e.g., a cylindrical, shape. In some embodiments, the transducer elements **204** are mounted movably and rotatably, providing mechanical degrees of freedom that can be exploited to improve focusing properties. Such movable transducers may be adjusted by conventional actuators, which may be driven by a component of controller **210** or by a separate mechanical controller.

Unlike many previous systems for focusing ultrasound through the skull, embodiments of the present invention utilize both longitudinal and shear waves. While water and soft brain tissue, for all practical purposes, only support longitudinal waves, shear waves can be excited in the much denser skull bone by longitudinal waves incident upon the skull surface. FIG. **3**A shows a CT image of a skull section **300**, in which the brightness level correlates with material density. The propagation of an ultrasound beam through the skull is illustrated schematically in FIG. **3**B, which shows an enlarged section of the skull **300**. A purely longitudinal wave L_I falling onto the outer skull surface **302** splits into a shear component S_{II} and a longitudinal component L_{II}. The relative contributions of these two components depend largely on the incidence angle. For wave frequencies between 100 kHz and 400 kHz, at angles below 20°, most of the energy of waves propagates in the longitudinal mode. At angles above 40°, the shear mode is predominant. In the intermediate regime, longitudinal and transverse modes co-exist. In general, part of the incident acoustic energy may be reflected (not shown). The remaining energy, which enters the skull, is partially absorbed, and partially propagated through the skull, depending on the frequency of the waves. For example, at frequencies above 500 kHz, most of the energy contained in the shear mode is absorbed and heats the skull. At frequencies between 100 kHz and 400 kHz, however, a significant fraction of the shear energy exits the skull at the interface **304** with the soft brain tissue, where it is transformed back into a longitudinal wave L_{IIIS}. The longitudinal wave L_{II} likewise penetrates the bone at such frequencies, and enters into the brain tissue as a longitudinal wave L_{IIIL}. The two components propagate through the skull **300** under different angles due to their different propagation velocities, and, consequently, enter the brain at different locations. In order to accurately adjust the amplitudes and phases of the transducer elements to focus the acoustic energy at the desired location, the energy contributions of both components are taken into consideration.

One approach to computing the propagation of longitudinal and shear waves through the skull involves numerically solving the Navier differential equation, which fully describes acoustic wave propagation, implying both longitudinal and transverse modes. Such a numerical simulation may be carried out, for example, using the finite elements method, and is therefore amenable to an arbitrary skull structure and geometry. It facilitates optimization of the phase and amplitude adjustments of the transducer elements for a particular patient, e.g., based on a CT scan of the patient's skull. However, this method is computationally expensive, and typically is performed off-line, i.e., during treatment planning.

Various alternative approaches are based on the analysis of the propagation of individual rays. For each transducer element, the path and phase shift of an acoustic ray starting at the element, passing through the skull bone, and arriving at the focal point is calculated. Using a simplified model of the skull which contains three locally parallel (e.g., concentric spherical) layers, corresponding to two cortical layers and a marrow layer in between, a closed analytical solution for the propagation of a ray through the skull may be obtained. Similarly, the propagation through other layers, e.g., soft brain tissue layers, may be calculated. The relations between incoming and outgoing waves for the various layers may be expressed in terms of transmission matrices, the product of which yields the overall transmission matrix for propagation from the transducer element to the focus. In this method, an acoustic ray is modeled as a planar wave, and includes both longitudinal and transverse modes.

A more realistic skull geometry can be accounted for in approaches that weight the contributions of longitudinal and shear modes, and calculate the propagation of these two modes separately. FIG. **4** illustrates separation methods in accordance with such approaches. In a first step **400**, an image of the patient's head may be taken, e.g., by magnetic resonance imaging or computer tomography. Step **400** is optional, as an otherwise obtained model of the skull and brain may be substituted for the image. For certain applications, a simplified model, such as a series of concentric spherical layers, may suffice to focus ultrasound with adequate accuracy. In step **402**, an array of transducer elements (see FIG. **2**) is positioned with respect to the skull, and the desired focus of the acoustic energy is specified. The information about the skull geometry and structure and about the locations of transducer array and focus are used repeatedly throughout the following computational steps, which may be carried out by controller **208**. First, an incident angle ϕ of the acoustic wave onto the skull is computed for each transducer element (step **404**). In an optional step **406**, the amount of energy reflected from the skull may be calculated based on the angle. The contributions of the transmitted energy to the longitudinal and the shear modes, respectively, are weighted in step **408**, as detailed further below. In step **410**, wave propagation through the skull and brain, and the phase shift and attenuation result-

ing at the focus, are calculated separately for each mode. Then, the contributions of the two waves are added to determine an overall phase shift and attenuation for waves emanating from the transducer element (step **412**). In embodiments wherein the transducer elements can be moved or rotated, they may then be adjusted to improve the focus (step **414**), and steps **404** through **412** may be repeated. Once the total phase and/or attenuation are determined for each transducer element, they are employed, in step **416**, to adjust the phase shifts α and amplification factors in the beamformer. Finally, the transducer array is driven accordingly to deliver acoustic energy to the desired focus (step **418**).

The weighting step **408** may be carried out in one of several ways of various levels of sophistication. FIGS. **5**A-**5**C illustrate three exemplary methods. In approach **408***a*, shown in FIG. **5**A, a threshold angle θ is determined (step **500**), and this angle separates a predominantly longitudinal regime from a predominantly transverse regime. The threshold angle may be chosen dependent on the frequency of the waves. For example, for acoustic waves at a frequency of around 0.2 MHz, a suitable threshold angle is 30°. The energy is assumed to be carried entirely by the longitudinal mode for incidence angles below the threshold angle, and entirely by the transverse mode for incidence angles above the threshold angle. Accordingly, the energy contribution of the shear wave is set to zero below the threshold angle (step **502**), and the energy contribution of the longitudinal pressure wave is set to zero above the threshold angle (step **504**).

FIG. **5**B illustrates a more involved method **408***b*, involving three regimes—a longitudinal regime, a shear regime, and an overlap regime. In step **510**, two threshold angles, θ_1 and θ_2, are defined. For incidence angles ϕ below the first, smaller threshold angle θ_1, the shear component is set to zero (step **512**), and for incidence angles above the second, larger threshold angle θ_2, the longitudinal component is set to zero (step **514**). At incidence angles between the first and second threshold angles, longitudinal and shear are assumed to coincide, and their relative contributions are determined by interpolation (step **516**). For example, the fraction of the shear component may be set to increase linearly from zero percent at the first threshold angle, θ_1, to one hundred percent at the second threshold angle, θ_2, as illustrated in the diagram of FIG. **6**. However, non-linear interpolations may also be used.

In some embodiments, illustrated in FIG. **5**C, the proportions of longitudinal and shear modes are determined empirically. Such methods **408***c* involve providing, in step **520**, values for the contributions of both modes over a range of incident angles, e.g., in the form of a graph. FIG. **7** is an exemplary graph showing the fraction of the incident energy that propagates through the skull in the longitudinal and transverse modes, based on empirically gathered data. At about 40°, the energy of the longitudinal mode drops to zero. The energy of the transverse mode increases from zero at normal incidence to as significant fraction at about 40°, above which angle of incidence it predominates. In step **522**, weighting factors for the longitudinal and the transverse component are determined for a particular incident angle based on, for example, the relationships shown in FIG. **7**.

With renewed reference to FIG. **2**, steps **404**-**412** are typically implemented by a suitable programmable processor and computer memory. The processor may be implemented in hardware or software, or a combination of both, within controller **208** or, alternatively, in a separate device such as a general-purpose computer. In addition, the operating program of system **200** may set aside portions of a computer's random access memory to provide control logic that effects one or more of the image capture, processing, and delivery of acoustic energy via transducer array **204**. In such an embodiment, the program may be written in any one of a number of high-level languages, such as FORTRAN, PASCAL, C, C++, C#, Java, Tcl, or BASIC. Further, the program can be written in a script, macro, or functionality embedded in commercially available software, such as EXCEL or VISUAL BASIC. Additionally, the software can be implemented in an assembly language directed to a microprocessor resident on a computer. For example, the software can be implemented in Intel 80x86 assembly language if it is configured to run on an IBM PC or PC clone. The software may be embedded on an article of manufacture including, but not limited to, "computer-readable program means" such as a floppy disk, a hard disk, an optical disk, a magnetic tape, a PROM, an EPROM, or CD-ROM.

Although the present invention has been described with reference to specific details, it is not intended that such details should be regarded as limitations upon the scope of the invention, except as and to the extent that they are included in the accompanying claims.

What is claimed is:

1. A method of delivering multimode ultrasound waves to a target through at least one tissue layer defining a first interface and a second interface, the waves having a shear component and a longitudinal component through the at least one tissue layer, the method comprising the steps of:

using a processor, assigning the waves to one of at least two regimes based at least in part on an incident angle of the waves at the first interface, the first regime corresponding to an incident angle smaller than a first threshold angle and the second regime corresponding to an incident angle equal to or greater than a second threshold angle;

using the processor, assigning weighting factors to the longitudinal and shear components of the waves based on the regimes to which the waves are assigned, the weighting factor of the shear component of a wave having a value of zero if the wave is assigned to the first regime and the weighting factor of the longitudinal component of a wave having a value of zero if the wave is assigned to the second regime;

using the processor, computing phase shifts associated with the shear component and the longitudinal component;

based on the weighting factors and the phase shifts, computing, using the processor, a weighted sum of the longitudinal and shear components so as to determine a total phase shift; and

compensating for the total phase shift-when delivering the ultrasound waves.

2. The method of claim **1** wherein the ultrasound waves are delivered by a phased array of transducers, the compensation step comprising adjusting outputs of the transducers in accordance with the total phase shifts determined from the weighted sums of the longitudinal and shear components associated with respective transducers.

3. The method of claim **1** further comprising computing coefficients of attenuation associated with the longitudinal component and the shear component, and compensating for the attenuation when delivering the ultrasound waves.

4. The method of claim **1** further comprising computing a coefficient of reflection of the wave at the first interface, and compensating for the reflection when delivering the ultrasound waves.

5. The method of claim **1** wherein assigning the waves to the regimes is further based on a frequency of the ultrasound waves.

6. The method of claim 1 wherein assigning the waves to the regimes is further based on an acoustic response of the first layer.

7. The method of claim 1 wherein the phase shifts associated with the shear component and the longitudinal component are computed based on at least one of a geometry, a thickness, a density, and an acoustic response of the first layer.

8. The method of claim 1 wherein the at least one tissue layer corresponds to a skull.

9. The method of claim 1 wherein the first threshold angle is equal to the second threshold angle.

10. The method of claim 1 wherein the waves are assigned to the first regime, the second regime or a third regime corresponding to an incident angle equal to or greater than the first threshold angle and smaller than the second threshold angle, and wherein determining weighting factors of the shear and longitudinal components comprises, if the waves are assigned to the third regime, interpolating between respective weighting factors for the first and second regimes.

11. The method of claim 10 wherein the interpolation is linear.

12. A method of delivering multimode ultrasound waves to a target through at least one tissue layer defining a first interface and a second interface, the waves having a shear component and a longitudinal component through the at least one tissue layer, the method comprising:

using a processor, computing phase shifts associated with the longitudinal and shear components;

based on an empirically determined relationship between relative energy contributions from the shear component and the longitudinal component and an incident angle of the waves at the first surface, assigning, using the processor, weighting factors to the longitudinal and shear components;

based on the weighting factors and the phase shifts, computing, using the processor, a weighted sum of the longitudinal and shear components so as to determine a total phase shift; and

compensating for the total phase when delivering the ultrasound waves.

13. The method of claim 1 further comprising providing a model of the at least one tissue layer, and basing the phase shift computation on the model.

14. The method of claim 13 wherein the model comprises locally parallel layers.

15. The method of claim 13 wherein the model is obtained using at least one of computer tomography or magnetic resonance imaging.

16. The method of claim 1 wherein the ultrasound waves have a frequency in the range from 100 kHz to 400 kHz.

17. The method of claim 1 wherein the ultrasound wave incident at the first interface is purely longitudinal.

18. The method of claim 17 wherein the ultrasound wave incident at the first interface excites a longitudinal component and a shear component thereat.

19. The method of claim 18 wherein the shear component excites a longitudinal wave at the second interface.

20. A system for delivering multimode ultrasound waves to a target located inside an object that comprises at least a first layer defining a first interface and a second interface, the waves having a shear component and a longitudinal component through at least the first layer, the system comprising:

(a) a phased array of transducers arranged around the object;

(b) a controller for

(i) receiving data about the object, its location relative to the transducers, and a location of the target,

(ii) computing for each transducer, based on the data, (1) an incident angle at which waves emanating from the transducer arrive at the first interface, and (2) phase shifts of the shear component and the longitudinal component,

(iii) for each transducer, computationally assigning the waves to one of at least two regimes based at least in part on the incidence angle, the first regime corresponding to an incident angle smaller than a first threshold angle and the second regime corresponding to an incident angle equal to or greater than a second threshold angle, and assigning weighting factors to the longitudinal and shear components based on the regimes to which the waves are assigned,

the weighting factor of the shear component of a wave having a value of zero if the wave is assigned to the first regime and the weighting factor of the longitudinal component of a wave having a value of zero if the wave is assigned to the second regime; and

(iv) based on the weighting factors and the phase shifts, computing a weighted sum of the longitudinal and shear components so as to determine a total phase shift for each transducer; and

(c) a beam former in communication with the controller and the phased array of transducers for adjusting outputs of the transducers to compensate for the total phase shift computed by the controller.

21. The system of claim 20 wherein the first threshold angle is equal to the second threshold angle.

22. The system of claim 20 wherein the at least two regimes comprise a third regime corresponding to an incident angle equal to or greater than the first threshold angle and smaller than the second threshold angle, and wherein the controller is configured to assign waves to the first regime, the second regime or the third regime based on the incident angles of the waves and to determine weighting factors of the shear and longitudinal components of waves assigned to the third regime by interpolating between respective weighting factors for the first and second regimes.

23. A system for delivering multimode ultrasound waves to a target located inside an object that comprises at least a first layer defining a first interface and a second interface, the waves having a shear component and a longitudinal component through at least the first layer, the system comprising:

(a) a phased array of transducers arranged around the object;

(b) a controller for

(i) receiving data about the object, its location relative to the transducers, and a location of the target,

(ii) computing for each transducer, based on the data, (1) an incident angle at which waves emanating from the transducer arrive at the first interface, and (2) phase shifts of the shear component and the longitudinal component,

(iii) for each transducer, computationally assigning weighting factors to the longitudinal and shear components based on an empirically determined relationship between relative energy contributions from the shear component and the longitudinal component and the incident angle of the waves at the first surface; and

(iv) based on the weighting factors and the phase shifts, computing a weighted sum of the longitudinal and shear components so as to determine a total phase shift for each transducer; and

(c) a beam former in communication with the controller and the phased array of transducers for adjusting outputs

of the transducers to compensate for the total phase shift computed by the controller.

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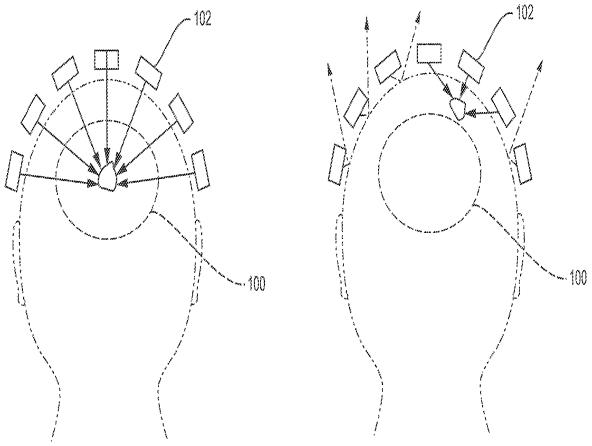
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